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Operative Versorgung bei Valgus und Varus Deformitäten - Optimierung der präoperativen Planung und intraoperativen Kontrolle anhand radiologischer Untersuchungen

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1. Affidavit

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Ich erkläre hiermit an Eides statt, dass ich die vorliegende Dissertation mit dem Titel:

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selbständig verfasst, mich außer der angegebenen keiner weiteren Hilfsmittel bedient und alle Erkenntnisse, die aus dem Schrifttum ganz oder annähernd übernommen sind, als solche kenntlich gemacht und nach ihrer Herkunft unter Bezeichnung der Fundstelle einzeln nachgewiesen habe.

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München, den 16.12.2024

Ort, Datum

Josef Brunner

Unterschrift Doktorand.

2. Inhaltsverzeichnis

3. List of abbreviations

AGA	German Association of Arthroscopy and Joint Surgery
AJC	Ankle joint centre
AP	anterior- posterior
CAD	Computer aided design
CAS	Computer assisted surgery
CT	Computed tomography
DFO	Distal femoral osteotomy
DGU	German Association of Trauma Surgery
DKG	German Association of Knee Surgery
DKOU	German Congress for Orthopedics and Trauma Surgery
DVT	Digital volume tomography
ESSKA	European Society of Sports Traumatology, Knee Surgery and Arthroscopy
HKA	Hip–knee–ankle angle
HTO	High tibial osteotomy
KA	kinematic alignment
KSSTA	(Journal for) Knee Surgery, Sports Traumatology, Arthroscopy
LLR	Long-leg radiograph
MA	Mechanical axis
MAD	Mechanical axis deviation
ML	Medial lateral
MLDFA	medial lateral distal femoral angle
MPTA	medial proximal tibial angle
MPFA	medial proximal femoral angle
MRI	Magnetic resonance imaging
MUM	Musculoskeletal University Centre Munich
owHTO	open wedge high tibial osteotomy
PSI	patient specific instrumentation
RAS	robotic assisted surgery
TKA	Total knee arthroplasty
TKC	Tibial knee centre

2D	Two-dimensional
3D	Three-dimensional

4. Publications

4.1 Publications of the cumulative dissertation

The following publications were written with my collaboration during the research activities in the Department of Orthopedics and Trauma Surgery in the Musculoskeletal University Center Munich (MUM).

The first two publications are part of this dissertation.

4.1.1 Paper 1

Significant changes in lower limb alignment due to flexion and rotation – a systematic 3D simulation study of radiographic measurements

Josef Brunner[1] and Maximilian Jörgens[1], Maximilian Weigert[2], Hannah Kümpel[2], Nikolaus Degen[1], Julian Fuermetz [1, 3]

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[The Paper was published in the journal: Knee Surgery, Sports Traumatology, Arthroscopy (KSSTA) an Official Journal of the European Society of Sports Traumatology, Knee Surgery and Arthroscopy (ESSKA)]
DOI: 10.1007/s00167-022-07302-x
Published: 3 Jan 2023
Shared first authorship with MD Maximilian Joergens

4.1.2 Paper 2

Linear correlation between patellar positioning and rotation of the lower limb in radiographic imaging: a 3D simulation study

Maximilian Jörgens[1], **Josef Brunner[1]**, Maximilian Weigert[2], Markus Bormann[1],Elisabeth Böhm[1], Wolfang Böcker[1], Alexander C. Paulus[1], Denis Ehrl[3], Julian Fürmetz[1, 4]

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[The Paper was published in the journal: Knee Surgery, Sports Traumatology, Arthroscopy (KSSTA) an Official Journal of the European Society of Sports Traumatology, Knee Surgery and Arthroscopy (ESSKA)]

DOI: 10.1007/s00167-023-07466-0
Published: 17 Jun 2023
Co-authorship

4.2 Publications additional to the cumulative dissertation

The following publication is **not** part of this dissertation, but should find appearance in this manuscript as it was also part of my work at the MUM. It underlines the relevance of the subject, as it is a good example for subsequent research projects based on our findings. The manuscript of the publication could be found in the appendix of this dissertation.

4.2.1 Paper 3

Open wedge high tibial osteotomy alters patellofemoral joint kinematics of the knee: a multibody simulation study

Lennart Schroeder[1], Sonja Ehreiser, **Josef Brunner[1]**, Klaus Radermacher[2], Boris Holzapfel[1], Maximilian Jörgens[1], Julian Fuermetz[1, 3]
[1] Department of Orthopaedics and Trauma Surgery, Musculoskeletal University Center Munich (MUM), University Hospital, LMU, Munich (Germany); [2] Chair of Medical Engineering, Helmholtz Institute for Biomedical Engineering, RWTH Aachen University, Aachen (Germany); [3] Department of Trauma Surgery, BG Unfallklinik Murnau, Murnau (Germany)

[Paper in review process: currently submitted to the journal “Journal of Orthopaedic Research”]
Co-authorship

4.2.2 Abstracts, posters and congress presentations

Furthermore, abstracts of the publications that contribute to this dissertation were presented and published in various congresses. In the following, they are listed in order of their chronological appearance.

Changes in lower limb alignment due to flexion and rotation – a 3D simulation of radiographic measurements

- *Jahreskongress Swiss Orthopedics – St. Gallen, CH, 06/2023 (poster)*
- *17. Deutscher Kongress für Orthopädie und Unfallchirurgie (DKOU) – Berlin, DE, 10/2022 (presentation)*
- *39. Kongress der Gesellschaft für Arthroskopie und Gelenkschirurgie (AGA) – Wien, AT, 09/2022 (poster/ presentation)*
- *11. Kongress der Deutschen Kniegesellschaft (DKG) – Leipzig, DE, 11/2021 (presentation)*

Bestimmung der Rotation in der Ganzbeinstandaufnahme anhand der Patellaposition

- *Jahreskongress Swiss Orthopedics – St. Gallen, CH, 06/2023 (poster)*
- *12. Kongress der Deutschen Kniegesellschaft (DKG) – München, DE, 11/2022 (poster)*
- *17. Deutscher Kongress für Orthopädie und Unfallchirurgie (DKOU) – Berlin, DE, 10/2022 (presentation)*
- *39. Kongress der Gesellschaft für Arthroskopie und Gelenkschirurgie (AGA) – Wien, AT, 09/2022 (presentation)*

The influence of open-wedge high tibial osteotomy on joint kinematics of the knee

- *Jahreskongress Swiss Orthopedics – St. Gallen, CH, 06/2023 (presentation)*
- *12. Kongress der Deutschen Kniegesellschaft (DKG) – München, DE, 11/2022 (presentation)*
- *17. Deutscher Kongress für Orthopädie und Unfallchirurgie (DKOU) – Berlin, DE, 10/2022 (presentation)*
- *39. Kongress der Gesellschaft für Arthroskopie und Gelenkschirurgie (AGA) – Wien, AT, 09/2022 (poster)*

4.2.3 Further book and journal contributions

Abstracts of our publications were published with kind permission of the KSSTA Journal

- **Signifikante Veränderungen des Alignments der unteren Extremität durch Flexion und Rotation – eine systematische 3D-Simulation**

  Brunner J, Jörgens M, Weigert M, Kümpel H, Degen N, Böcker W, Fürmetz
  Spitzenforschung in der Orthopädie und Unfallchirurgie. Innovationen und Auszeichnungen 2023,
  hrsg. von der ALPHA Informations- GmbH, Lampertheim 2023, S. 32 – 37

- **Veränderungen in der Ausrichtung des Beins durch Flexion und Rotation**

  Brunner, J, Jörgens, M, Weigert, M, Kümpel H, Fürmetz J
  Arthroskopie (2023). https://doi.org/10.1007/s00142-023-00597-z

4.3 Prices and Awards

An Abstract of the publication “Significant changes in lower limb alignment due to rotation and flexion – a systematic 3D simulation study of radiographic measurements” was honored with two different awards.

- **Innovationspreis der Deutschen Gesellschaft für Unfallchirurgie (2022)**

  17.Deutscher Kongress für Orthopädie und Unfallchirurgie (DKOU)
  Berlin, DE, 10/2022
  (10.000€)

- **1. Poster Preis der Gesellschaft für Arthroskopie und Gelenkschirurgie (2022)**

  39. Kongress der Gesellschaft für Arthroskopie und Gelenkschirurgie (AGA)
  Wien, AT, 09/2022
  (1.000€)

5. Introduction

5.1 General introductionist

Over the past few decades, incidence, quality, and complexity of deformity correction surgery increased continuously. With a total of around 172.000 annually performed total knee arthroplasties (TKA), and 28.300 correcting osteotomies such as open wedge high tibial osteotomy (HTO) and distal-femoral-osteotomies (DFO), they represent some of the most frequently applied orthopedic interventions in Germany (1). Even though the majority of TKA operations have demonstrated promising long-term results with outstanding survival rates of the implants of up to 82% after 25 years, not less than 20% of the patients remain unsatisfied with the clinical outcome after one year (2-5).

Due to numerous influencing factors, heterogeneous register data and continuous development of surgical techniques, it seems almost impossible to identify isolated reasons for dissatisfaction and less successful outcomes. But, as implant malalignment has been reported as an important contributing factor to the development and progression of osteoarthritis, tremendous work has been invested into the optimization of implant design and positioning (6-8). Especially computer assisted surgery (CAS), using robotic arms and patient specific instrumentation (PSI) are important technological inventions for achieving unrivalled accuracy of the implantation with direct feedback on the implant positioning. Even though, CAS is constantly subject of scientific publications and various robotic assisted surgery (RAS) systems are ordinarily used in some clinics, the integration into daily clinical practice remains unaccomplished (9).

In course of these improvements in navigation and implant positioning, the aim for an optimal alignment method of the implant has characterized recent research ambitions.

Although there has been no absolute consensus about the ideal alignment method, there is a tendency towards favoring the kinematic alignment (KA) method, which is orientated at the patient`s individual anatomy according to ligamental and muscular forces (10-12). According to the latest scientific discourse, this approach has been identified with a better clinical and functional outcome for the patient and a comparably long survival of the prosthesis (10-16). The kinematic alignment of individualized prosthesis (in TKA`s) and cutting blocks (in HTO`s) is intended to be within a precision range of 3-4° variance, since kinematic alignment with slight varus angulation is associated with better clinical outcome. In contrast, alignment with additional alteration of more than 3° is associated with asymmetric weight bearing, advanced implant aberration, early insufficiency and arthrosis (17). As a consequence, the precision and reliability of limb alignment measurements need to be within this reported range.

Regardless of the chosen alignment method, malalignment in general leads to the development and progression of osteoarthritis. There is also a high evidence for a correlation between surgical inaccuracy of deformity correction and inferior clinical outcome with early conversion into TKA (18, 19).

So, two of the most important prerequisites for an optimal surgical outcome of both osteotomies and

TKA`s are an accurate preoperative planning procedure and second a reliable postoperative evaluation using radiological examinations (20).

Notwithstanding the availability of three-dimensional (3D) imaging techniques such as magnetic resonance imaging (MRI), computed tomography (CT), EOS® 2D/3D imaging and digital volume tomography (DVT), these examinations are still commonly performed using two-dimensional (2D) long-leg radiographs (LLR) with subsequent manual dimensioning and measurements (20, 21).

5.2 Long Leg Radiographs

The major advantage of this imaging modality is a precise assessment of the mechanical axis of the entire lower limb, which is of particular importance while grading a kinematic alignment successful or not (22). As various studies stated, that the LLR alone provides accurate information on mechanical axis measurements when lower limb malalignment was suspected (23, 24).

Furthermore, postoperative LLR`s after both osteotomies and TKA`s, can be compared with preoperative images and in reference with standard values for lower limb alignment established over decades (20, 25).

For many years, the initial definition of D. Paley served as the standard observation procedure for LLR`s (21). For a "neutral position", it should be obtained in upright standing position, with fully extended knees and in true anterior-posterior (AP) view of the knee with the patella centered between the femoral condyles (26). In many cases of axial deformities, it is necessary to rotate the tibiofibular compartment to achieve a position with centralized patella in the coronal plane (27). Alongside frequent rotational variance in lower limb orientation, extension deficits that often occur along symptomatic osteoarthritis, axial deformities, TKA or other causes of partial immobility, lead to undesired variable examination conditions with changing patient positions (28-30). A reduced range of motion in the hip joint due to femoroacetabular impingement or degenerative changes, complex underlying bone deformities and patella malformations also tribute to heterogeneous examination conditions (31, 32). According to Ritter et al., who examined over 5600 cases of TKA`s over multiple decades, one third of all patients undergoing TKA surgery had a severe preoperative flexion contracture between 6° and 50° (33).

Also, an absolute consensus concerning the correct "neutral position" of the lower limb during image acquisition and potential alternatives to a centralized patella, remains unestablished (18, 28, 34). Besides various approaches, some authors describe and suggest a "knee forward" orientation, with the femur condyles orthograde to the sagittal axis, parallel with the frontal plane, and tangential to the radiographic detector plane (35). This modality was hoped to be less influenced by present patellofemoral malalignment and tibial torsion (27, 35). The comparison between these two modalities regarding their accuracy and suitability in a clinical context is part of this investigation.

In conclusion, many LLR`s are recorded in positions with flexional and rotational impact, which influence the alignment of the mechanical axis and various angles significantly (20).

5.3 Position dependencies and clinical relevance

Undoubtedly, there are several other modalities like the biplanar linear radiograph system (EOS®), that would better depict the 3D joint anatomy or even allow 3D reconstructions with potentially lower rates of malrotated limbs during image acquisition (36-40). But as long as those alternatives are not routinely available in most clinics, weight-bearing LLR`s persist as gold standard for alignment assessment (28). In some countries like Germany, LLR`s are even mandatory for specialized centers to get certified (41).

To date, there is no automated assessment of LLR`s, that takes malrotation and flexion as well as deviations of the patient`s position from pre- to postoperative images into account (20). Therefore it is even more important for clinicians to be aware of potential incorrect measurements and subsequent unprecise surgical planning (17, 29, 30, 42). Unless, the influences of either rotation or flexion on some of the common radiographic alignment parameters were already investigated by several studies using synthetic bone models and 3D simulation programs, there were only two studies that examined combined effects of rotation and flexion, using sawbone models, cadaveric and in vivo studies with very small cohorts or even singular probands (17, 29, 30, 43-46).

Despite a large heterogeneity regarding their methods, cohorts and explicit results, they all conclude that malrotation of the lower limb is present in many LLR`s and alters the mechanical alignment significantly. For example, the HKA showed a mean change of > 2° between 15° internal and 15° external rotation in every study that was analyzed by a major systematic review of Ahrend et al. (28). Unless these effects were considered to be small, they are within the range that advanced surgeons can achieve (47). This separates a successful from an insufficient kinematic alignment resulting in over- or undercorrection and poor postoperative outcome (33, 48). Furthermore, it was consensually hypothesized, that the impact of rotation would reach high clinical importance when additional sagittal knee flexion was present (17, 28-30, 49). With incremental rotation and flexion of the limb, also a precise assessment of the patella position and detection of patella malformations is challenging (49, 50).

Therefore, the use of LLR`s for accurate surgical planning in case of severe deformities or acute injuries, is highly discussable, when standardized positioning is not possible (20). The repetition of LLR`s, that has been identified with malrotation, seems like a logical option, but a higher cumulative radiational exposure as a clinical consequence must not be neglected.

5.4 Study goals

One main goal of this study was to investigate the suspected position dependencies of the mechanical axis and joint angles of the lower extremity due to flexion and rotation, which were regarded as much greater than singular effects (20, 27, 34).

We further aimed to systematically merge together the focuses and heterogenous methods of former studies by using a 3D simulation program and to establish a possibility to easily quantify effects for a larger cohort (20). The Geomagic simulation program enabled us to project angle measurements into the coronal plane to imitate LLR`s, apply rotation and flexion with high accuracy and refer all models to an exact coordinate system to guarantee comparability between the patients (20, 27).

With the knowledge of how strong combined rotation and flexion would alter common limb alignment parameters, the next step was to investigate a possible correlation between the degree of rotation with changes in joint angles and the resulting position of the patella (27). Background to this investigation was the call for a suitable clinical tool to predict underlying rotational and flexional influences, just by assessing the relative patellar position changes between image pairs. There was already an attempt to calculate underlying position changes based on tibiofibular overlap, but not yet on patellar positioning (27, 51). Thus, we aimed to scrutinize the postulated direct correlation between rotation of the limb and patella position. Additionally, we quantified the necessary underlying rotation that would lead to a centralized patella and examined whether a “knee forward” orientation would be a better acquisition position for LLR`s. We postulated this modality to be less influenced by present patellofemoral malalignment and tibial torsion than D. Paley’s procedure (28, 47).

In the third publication, that is currently under revision, we used the Geomagic Software to simulate owHTO interventions with various open tibial wedges as well as a Python script to investigate resulting alteration of mechanical angles and distances. In combination with a virtual musculoskeletal body modeling system, we aimed to gain extensive comprehension of the impact, that changes in lower limb alignment due to owHTO`s have on knee joint kinematics (see appendix).
We consider our findings as a further step in improving the reliability of postoperative accuracy examination and the clinical outcome of osteotomies in general.

Summarizing all mentioned study ambitions, the long-term goal of the project is to contribute to the optimization of an automated 3D planning software that enables improved planning and reliable measurement of the precision of knee operations.

5.5 Contribution to the conduction of this work

Both publications presented in this dissertation are a result of the collaboration between members of the Musculoskeletal University Centre Munich (MUM) and the Statistical Consulting Unit StaBLab of the LMU Munich.

5.5.1 Contribution to Paper 1

In the following, I shortly describe my personal contribution to the conduction of Paper 1, *“Significant changes of lower limb alignment due to flexion and rotation - a systematic 3D simulation study of radiographic measurements”*

As mentioned in the publication, “JB [Josef Brunner] and MJ [Maximilian Jörgens] were responsible for conceptualization, methodology, investigation, software, formal analysis, data curation, visualization, funding acquisition and writing the original draft” (20)

First of all, formulating the working hypothesis that LLR`s are prone to error due to underestimated underlying rotational and flexional influences on common alignment parameters was formulated and a suitable study design was conceptualized. After extensive literature recherche, prerequisites for the subsequent simulation were organized. Getting into touch with the Geomagic simulation software and evaluating the suitability of preexisting 3D bone models derived from former studies regarding their completeness, were crucial in the beginning (25). Next, the implementation of validated and clinically relevant landmarks on the surface of the models was necessary (25). To refer position changes of the models to a neutral position, a reference coordinate system was integrated into the virtual environment of every model (52). Conceptualization of the movements required to imitate flexion and rotation of the limb in accordance with biomechanical considerations took place, after consulting the biomechanical research lab of the LMU.

One of the biggest work steps was the manual simulation of flexion and rotation for all 60 virtual models. Due to segmentation of the virtual legs derived from CT data, every single part of the limb must have been rotated and flexed separately.

In total 8400 individual position changes were necessary, to simulate internal and external rotation up to ±15° and flexion up to 30° for 60 virtual bone models. After first simulations, preliminary results were controlled regarding their plausibility and in cases of incorrect simulations they were repeated. The tremendous workload of manual simulation called for the design and implementation of an automated solution for generating the simulation data. So, I adapted and extended a Python script that enabled us the iterative calculation of mechanical axes, angles and distances between our landmarks and made manual simulations obsolete. In addition a Python code to project three dimensional angular measurements into the coronal plane to mimic radiographic imaging, was programmed.

After the simulations, the raw data output from the Python script was translated into a structured Excel form to enable descriptive statistical analysis. We performed the statistical analysis in cooperation with the Statistical Unit of the LMU Munich (StaBLab).

For the graphical illustration of the results and visualization of the models, coordinate system and the simulation, I used simple visualization tools within the Geomagic software and Adobe Photoshop for rendering. After all I was mainly responsible for writing the original draft and revising the manuscript together with the co-authors.

Adaptions, translations and extensions of the manuscript such as posters and abstracts for presentation at congresses were made.

In this publication I share the first authorship with Maximilian Jörgens who was a crucial contributor to the conceptualization of this work. We collaborated immensely in implementing, adapting, examining and testing the simulation regarding its suitability for the study and finding solutions for upcoming problems. Furthermore, he wrote parts of the original draft, was responsible for corrections and finalization of the manuscript. Therefore, we assumed it legit sharing the first authorship.

5.5.2 Contribution to Paper 2

In the following, I shortly describe my contribution to the conduction of Paper 2, *“Linear correlation between patellar positioning and rotation of the lower limb: a 3D simulation study”*

As mentioned in the manuscript, “MJ and JB were responsible for methodology, investigation, formal analysis, data curation, writing the original draft and visualization” (27).

After the methodical work for the first study has been finished, the conceptualization and formulation of the working hypothesis for the second study was done.

Besides an extensive literature research to explore the current state of knowledge in the specific research field, further required landmarks around the patella were implemented into the virtual bone models. A method for calculating the patellar position in relation to the tibia with anatomical landmarks recruitable from LLR`s needed to be established. Furthermore, a Python code to calculate changes of the patella simultaneously with the simulation, was written. I did a manual simulation of lower limb rotation in 5 degree steps and later on in 1 degree steps in the same manner as in the previous study and calculated the changes in alignment parameters (27). Some improvements of the formula regarding biomechanical considerations, have been made. In line with ambitions towards a more automatized workflow, I have designed and implemented an algorithm that could simulate any required movements within the virtual model without any interactive simulation using Geomagic (27). After data transfer into an Excel compatible format, I analyzed the results in a descriptive manner and assessed their plausibility. I also did the visualization and graphical illustration of the methods and results and was responsible for writing the original draft.

5.5.3 Contribution to Paper 3

The third paper *“Open wedge high tibial osteotomy (owHTO) shows relevant impact on tibiofemoral and patellofemoral joint kinematics of the knee in a multibody simulation model”* is not part of this dissertation. It is added for better understanding of the general research ambitions of the entire project.

As mentioned in the manuscript “ JB was responsible for data curation, review and editing” [Quelle Schröder], my contribution to this project was data acquisition, writing parts of the manuscript and reviewing. Mainly, I simulated multiple open wedge high tibial osteotomies using the same Geomagic simulation software as in the previous studies. After the manual simulation and simultaneous data generation with a slightly adapted Python script to measure alignment parameters, I transferred the data into a Excel compatible format and curated them for further analysis. Additionally, I was responsible for writing parts of the methods chapter and editing the original draft.

6. Summary

In contrast to the rising popularity of 3D imaging modalities such as MRI, DVT, EOS and CT, most of the preoperative planning procedures and postoperative evaluations in deformity correcting surgery of the lower limb are still commonly performed using 2D radiographic imaging (53, 54). For many years, the LLR obtained with a centralized patella, was considered to be the optimal prerequisite for interventions as HTO`s and TKA`s, as it allows a standardized and simple image acquisition. Furthermore, well established norm values for alignment parameters and the possibility of a sensitive identification of mechanical axes (MA), made LLR`s crucial for a reliable postoperative assessment (20, 28, 55). Previous experiments showed remarkable dependencies of lower limb alignment measurements on the patient`s position and emphasized the importance of a correct and ever standardized image acquisition (17, 29, 30, 43, 45, 49).

We investigated the singular and combined effects of rotation and flexion on various established mechanical alignment parameters using virtual bone models derived from CT data. After the implementation of a coordinate system, sixty models were each manually rotated around the longitudinal axis and flexed along the intercondylar axis in incremental steps up to 15° respectively 30° (27). To mimic radiographic imaging, 3D joint angles were projected into the coronal plane. Huge effort was put into the automatization of simultaneous angular measuring by programming a Python based algorithm, that luckily could been used in multiple simulations (20, 27). Following biomechanical considerations, we hypothesized that the combined effect of rotation and flexion might alter alignment greater than singular effects (20). The observed data revealed small effects for isolated rotation or flexion, but pronounced and clinically relevant alteration when they were combined. For example, the MAD showed ranges of ±25mm variation in relation to the physiological norm values described by D. Paley, when 15° rotation were combined with 30° flexion. A similarly strong reaction to combined influences was presented by the HKA angle with variations of 0.03° per degree limb rotation with extended knee, but with up to 0.6° per degree when additional 30° flexion was present (20). The mechanical lateral distal femoral angle (mLDFA) and the medial proximal tibial angle (MPTA) displayed comparably concise alterations (20).

Thus, the need for an easy applicable clinical tool to predict underlying malrotation in pre- and postoperative image pairs has been strong (51). In line with this, we investigated lower limb rotation up to 15°, subsequent changes in angular measurement and patella position concerning a potential intrinsic correlation. In conclusion, an approximately linear relationship with a − 0.9 mm change of patella position per degree could be seen, making an inverse estimation of rotation more likely (27). In accordance with our findings, it seems feasible for clinicians to estimate present malrotation and corresponding alignment parameters by looking at the patella position. An implication of the formula into an automatized recognition algorithm, like Maderbacher et al. did it with the method of tibia fibula overlap assessment, remains part of future investigations.

Although it was not of primal interest, we scrutinized the differences in alignment between image pairs, one with a centralized patella and one with orthograde positioned condyles (27). An average internal rotation of − 9.8° must have been applied before a centralized patella was achieved. While merging the findings of our twofold study, it can be assumed that this commonly present rotation impacts the alignment parameters more than the reported tolerance of 3° and kinematic alignment based on these radiographs would be prone to error (20, 47).

In conclusion, the impact of rotation and flexion on alignment parameters calls into question the current gold standard of obtaining LLR´s with a centered patella as a prerequisite for excellent surgery (27, 35).

7. Summary in German (Zusammenfassung)

Im Gegensatz zur zunehmenden Beliebtheit von 3D-Bildgebungsmodalitäten wie MRT, DVT, EOS und CT wird ein Großteil der präoperativen Planungsverfahren und postoperativen Evaluierungen in der Deformitätenchirurgie der unteren Extremität immer noch standardmäßig anhand von 2D-Röntgenaufnahmen durchgeführt (53, 54). Viele Jahre lang galt die Ganzbeinstandaufnahme (GBSA), aufgenommen mit zentralisierter Patella, als optimale Aufnahmemodalität bei Eingriffen wie Korrekturosteotomien und Kniegelenksprothesenimplantationen, da sie bei Durchführung nach standardisiertem Procedere mit etablieren Normwerten vieler Winkel und Streckenbeziehungen aufwartet (20, 25, 56). Darüber hinaus ist die Möglichkeit einer sensitiven Identifizierung mechanischer Achsen (MA) in der GBSA, eine entscheidende Voraussetzung für eine zuverlässige postoperative Beurteilung (20, 28). Jüngste Experimente zeigten jedoch eine ausgeprägte Positionsabhängigkeit vieler Messgrößen in der zweidimensionalen Röntgenaufnahme und betonten die Bedeutung einer korrekten und stets standardisierten Bildaufnahme als Grundvoraussetzung für ein optimales Operationsergebnis (17, 29, 30, 34, 43). Insbesondere klinisch bedingte Streckdefizite und Rotationseinflüsse scheinen die Bemaßung mechanischer Achsparameter signifikant zu beeinflussen. Auf Basis biomechanischer Überlegungen und ausführlicher Literaturrecherche stellten wir die Hypothese auf, dass der kombinierte Effekt von Rotation und Flexion die Ausrichtung stärker verändern würde als die einzelnen Effekte (20).

Deshalb untersuchten wir die individuellen und kombinierten Effekte von Rotation und Flexion auf verschiedene etablierte mechanische Ausrichtungsparameter anhand von virtuellen Knochenmodellen, die aus CT-Daten generiert wurden (20, 27). Nach der Implementierung eines Referenzkoordinatensystems wurden 60 Modelle jeweils manuell um die Längsachse um bis zu 15° gedreht und entlang der interkondylären Achse zunehmend bis 30° gebeugt (20). Um eine Röntgenaufnahme zu imitieren, wurden die 3D-Gelenkswinkel in die Koronarebene projiziert und anschließend systematisch im Bezug zu den etablierten Normwerten analysiert.

Um eine simultan neben der Simulation ablaufende Winkelmessung zu ermöglichen, wurde beträchtlicher Aufwand in die Entwicklung eines Algorithmus für die Automatisierung der Messung gesteckt. Dieser auf einem Python Script basierende Algorithmus dient als Grundlage für die Winkelmessungen aller in dieser Arbeit erwähnten Untersuchungen. In der Zusammenschau der beobachteten Daten ließ sich unsere ursprüngliche Hypothese, wonach sich geringe Auswirkungen bei isolierter Rotation oder Flexion, aber ausgeprägte und klinisch relevante Veränderungen bei Kombination zeigen würden, für alle untersuchten Messgrößen bestätigen (20).

Die mechanische Achsabweichung (MAD) beispielsweise zeigte Schwankungsbreiten von ±25 mm in Relation zu den mittleren Ausgangswerten, welche sich in dem von D. Paley definierten Normbereich befanden (20, 26).

Mit einer ähnlich starken Reaktion auf kombinierte Einflüsse zeigte der HKA-Winkel Veränderungen von 0.03° pro Grad Rotation bei gestrecktem Bein, aber bereits 0.6° Abweichung pro Grad Rotation, wenn eine zusätzliche Flexion von 30° vorlag (20). Vergleichbar prägnant waren auch die Veränderungen des mechanischen lateralen distalen femoralen Winkels (mLDFA) und des medialen proximalen tibialen Winkels (MPTA).

Im Zuge des zunehmenden Einsatzes von Planungssoftware, die mithilfe intelligenter Bilderkennung von Landmarken, eben erwähnte Winkel automatisch berechnen, sind unsere Ergebnisse dringend zu beachten (25, 57, 58).

Daher sehen wir großen Bedarf an einem einfach anwendbaren klinischen Instrument zur Vorhersage der zugrunde liegenden Fehlrotation im Vergleich von prä- und postoperativen Bildpaaren. Wir untersuchten bei zunehmender Rotation bis 15° die resultierenden Veränderung der Winkel und der Position der Patella hinsichtlich einer möglichen direkten Korrelation (27). Es zeigte sich ein annähernd linearer Zusammenhang mit einer Veränderung der Patellaposition um - 0,9 mm pro Grad, was eine inverse Schätzung der Rotation im dreidimensionalen Raum in den Bereich des Möglichen rückt (27). Unter Berücksichtigung unserer Ergebnissen scheint es für Kliniker machbar zu sein, eine vorliegende ungewollte Rotation sowie die entsprechenden Ausrichtungsparameter anhand der Patellaposition abzuschätzen. Eine klinische Umsetzung mit dem Ziel einer einfachen Formel beziehungsweise eines Algorithmus, ähnlich wie es die Kollegen Maderbacher et al. anhand des Tibia Fibula Overlaps vollzogen haben, ist ein mögliches Ziel für die Zukunft (51).

Obwohl, gemäß Literatur, eine Zentrierung der Patella bei GBSA vorausgesetzt wird, besteht bezüglich der Vorteile im Gegensatz zu Aufnahmen mit orthograd positionierten Femurkondylen kein eindeutiger Konsens. Anhand unserer Simulationsergebnisse konnten wir zeigen, dass eine durchschnittliche Innenrotation von - 9,8° zur Zentrierung der Patella durchgeführt werden musste (27). Während unsere Simulation anhand von Modellen mit physiologischen Patellapositionen durchgeführt wurden, ist davon auszugehen, dass dieser Effekt bei Patella Malformationen noch ausgeprägter sein wird.

Die Auswirkung dieser erforderlichen Rotation auf die Ausrichtungsparameter ist noch nicht vollständig geklärt, stellt aber die strenge Ausrichtung der GSBA nach zentralisierter Patella mindestens in Frage (27, 35, 59).

8. Paper I

Significant changes in lower limb alignment due to flexion and rotation – a systematic 3D simulation study of radiographic measurements

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Significant changes in lower limb alignment due to flexion and rotation—a systematic 3D simulation of radiographic measurements

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Abstract

Background Many radiographic lower limb alignment measurements are dependent on patients' position, which makes a standardised image acquisition of long-leg radiographs (LLRs) essential for valid measurements. The purpose of this study was to investigate the influence of rotation and flexion of the lower limb on common radiological alignment parameters using three-dimensional (3D) simulation.
Methods Joint angles and alignment parameters of 3D lower limb bone models (n = 60), generated from computed tomography (CT) scans, were assessed and projected into the coronal plane to mimic radiographic imaging. Bone models were subsequently rotated around the longitudinal mechanical axis up to 15° inward/outward and additionally flexed along the femoral intercondylar axis up to 30°. This resulted in 28 combinations of rotation and flexion for each leg. The results were statistically analysed on a descriptive level and using a linear mixed effects model.
Results A total of 1680 simulations were performed. Mechanical axis deviation (MAD) revealed a medial deviation with increasing internal rotation and a lateral deviation with increasing external rotation. This effect increased significantly ($p < 0.05$) with combined flexion up to 30° flexion (− 25.4 mm to 25.2 mm). With the knee extended, the mean deviation of hip–knee–ankle angle (HKA) was small over all rotational steps but increased toward more varus/valgus when combined with flexion (8.4° to − 8.5°). Rotation alone changed the medial proximal tibial angle (MPTA) and the mechanical lateral distal femoral angle (mLDFA) in opposite directions, and the effects increased significantly ($p < 0.05$) when flexion was present. **Conclusions** Axial rotation and flexion of the 3D lower limb has a huge impact on the projected two-dimensional alignment measurements in the coronal plane. The observed effects were small for isolated rotation or flexion, but became pronounced and clinically relevant when there was a combination of both. This must be considered when evaluating X-ray images. Extension deficits of the knee make LLR prone to error and this calls into question direct postoperative alignment controls. **Level of evidence** III (retrospective cohort study).

Keywords 3D simulation · Radiographic measurement · Coronal alignment · Lower limb rotation · Knee flexion

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Abbreviations

AJC	Ankle joint centre
CT	Computed tomography
DVT	Digital volume tomography
FHC	Femoral head centre
FNP	Femoral notch point
HKA	Hip–knee–ankle angle
aLDFA	Anatomic lateral distal femoral angle
LPFA	Lateral proximal femoral angle
LLR	Long-leg radiograph
wbLLR	Weight-bearing long-leg radiograph
MA	Mechanical axis
MAD	Mechanical axis deviation

MFA Mechanical femoral axis
MTA Mechanical tibial axis
ML Medial lateral
mLDFA Mechanical lateral distal femoral angle
MPFA Medial proximal femoral angle
MPTA Medial proximal tibial angle
MRI Magnetic resonance imaging NSA
Neck shaft angle
TKA Total knee arthroplasty
TKC Tibial knee centre
2D Two-dimensional
3D Three-dimensional

Introduction

Despite the availability of three-dimensional (3D) imaging techniques such as magnetic resonance imaging (MRI), computed tomography (CT), EOS® 2D/3D imaging, and digital volume tomography (DVT), preoperative surgical planning is still commonly performed on two-dimensional (2D) long-leg radiographs (LLRs) [5]. The main advantages are standardized, fast and easy image acquisition, as well as broad availability, with standard values for lower limb alignment established over decades [21]. Furthermore, LLRs can identify anatomic variations of the femur and the tibia with high sensitivity by easily assessing the mechanical axis [21]. Additionally, intraoperative fluoroscopic images can be compared with these preoperative images [9, 23].

The standardized observation procedure of LLRs is in upright standing position, with the knee fully extended and a centralised patella in the frontal plane [21]. Many patients with axial deformities, osteoarthritis, total knee arthroplasty (TKA) or other causes of partial immobility cannot fully extend their knees. Yet, 2D X-ray projection images change depending on the patient's position and are influenced by rotation and flexion [1, 5, 8, 11, 25].

This leads to difficulties in reliably performing LLRs with limited comparability of pre- and postoperative images [2]. The use of LLR is, therefore, questionable for accurate surgical planning in cases of severe deformities or acute injuries, where standardized positioning is not possible [10, 15].

Several studies have investigated the influence of either rotation or flexion on lower limb alignment measurements, and two studies examined combined effects on some of the common radiographic alignment parameters, using a syn- thetic bone model and 3D simulation programs. Following biomechanical and kinematic considerations, these com- bined effects were considered to be much greater than those of rotation or flexion alone [10, 13].

Therefore, there was an urgent need to investigate how strong the combined effects were within a larger population and so the focuses of several studies on different clinically important mechanical measures [HKA (hip–knee–ankle angle), MPTA (medial proximal tibial angle), mLDFA (mechanical lateral distal femoral angle), MAD (mechanical axis deviation)] were systematically merged together in a comprehensive manner [11, 13, 16, 23].

To date, there is no study yet, that examined these postulated combined effects due to rotation and flexion on various established mechanical alignment parameters based on virtual CT models. The aim of this study was to quantify the influence of combined rotation and flexion of the lower limb on common alignment parameters using 3D simulation. Based on biomechanical and kinematic considerations, combined effects were assumed to be much greater than from rotation or flexion alone.

Materials and methods

For this software and program-based simulation study, 60 3D bone models of the lower limb were used, that were created from existing anonymized CT-data of 30 randomly selected patients (18–50 years) showing alignment parameters within the range of reported norm values and indicating the absence of any severe deformity in coronal neutral position (Table 1) [21]. To cover side differences between left and right limbs, both sides of each of the 30 patients were included. Exclusion criteria were advanced osteoarthritis of the hip joint and knee joint, radiographic evidence of previous realignment surgery, fractures, any lower extremity joint replacement, and age above 50 years. Physiological homogeneity of the selected patient collective was chosen to test the hypothesis, before deformities and more variable coronal alignment could be investigated [7, 19]. Digital 3D copies were processed using the validated rendering soft- ware program, Mimics 14.0 (Materialize, Leuven, Belgium), for segmentation and calculation of the CT images and subsequently using the Geomagic Studio 2014 (3D Systems, Morrisville, NC, USA) software to create a 3D geometry of the leg [5]. A standardized new coordinate system was set

Table 1 Summary of alignment measurements of the models for simulation (n = 60); *HKA* hip–knee–ankle angle, *MPTA* medial proximal tibial angle, *MAD* mechanical axis deviation, *mLDFA* mechanical lateral distal femoral angle

	HKA (in °)	MPTA (in °)	mLDFA (in °)	MAD (in mm)
Mean	180.1	87.7	87.2	6.2
Minimum	171.3	82.3	83.1	−11.5
Maximum	187.7	92.7	92.9	28.8
Standard deviation (SD)	±3.1	±2.6	±2.2	±8.4

in every model and enabled us to relate positional changes due to flexion and rotation back to the physiological neutral position (Fig. 1). According to the methods of Miranda et al. the coordinate system was implemented based on mechanical axes, principal mass and cylindrical surface fitting [17]. Considering the need for accurate measurements, a method that uses coordinates established with high accuracy and reliability in previous publications, was chosen [5].

Definition of angles and points

As it was aimed to quantify changes in angular measurements, validated and publication-based 3D landmarks were integrated into all models [5]. Their projection into the coronal plane finds an approximate equivalent to the 2D landmarks of D. Paley, which are commonly used in this research field [5, 21]. A python code was written in which the y-coordinate was set to zero and all measurements were automatically projected into the coronal plane to mimic radiographic imaging [5, 16, 23].To evaluate the changes in alignment, after every simulation step, angles and distances were automatically evaluated with another python script, and measured results were statistically analysed. By convention, negative measurements indicated the lower extremity to be internally rotated and positive measurements externally rotated around the longitudinal mechanical axis [8].

The centre of the femoral head (FHC), the femoral notch point (FNP) and the centre of the tibial articular surface of the ankle joint (AJC) were chosen to define the mechanical axis (MA) [8, 21]. According to the study by Moreland et al., the current study utilized the FNP as the femoral centre of the knee and the centre between the tibial spines on the tibial surface as the tibial centre of the knee (TKC) [5, 18]. To measure the HKA, the connecting lines between FHC and FNP, as well as between TKC and AJC were created. This angle is defined as the medial angle between those two vectors [10, 21]. MAD was calculated as the distance of the MA from the centre of the knee joint. The most distal points of the femoral condyles and the most proximal lateral and medial points of the tibia were necessary to describe mLDFA and MPTA [5, 8, 21].

Simulation of flexion and rotation

The models were then aligned to the new coordinate system and a neutral origin position (0° flexion, 0° rotation) was set. Next, all models were rotated around the longitudinal MA in 5° increments up to 15° internally and 15° externally and additionally flexed in 10° steps along the femoral transepicondylar axis up to 30° (Fig. 2). For every model, 28 combinations of flexion and rotation were simulated, which led to 1680 positions in total.

Half of the flexion was performed on the femoral part of the model and half on the tibial part in reverse direction. The division of motion and the determination of the vertical long axis from the FHC to the ankle joint were performed according to the methods of Jud et al. to obtain a realistic position compared with radiographs [10]. Furthermore, the screw home motion of the knee joint in the last 20° of extension by additionally internally rotating the tibia 5° during flexion was simulated [24].

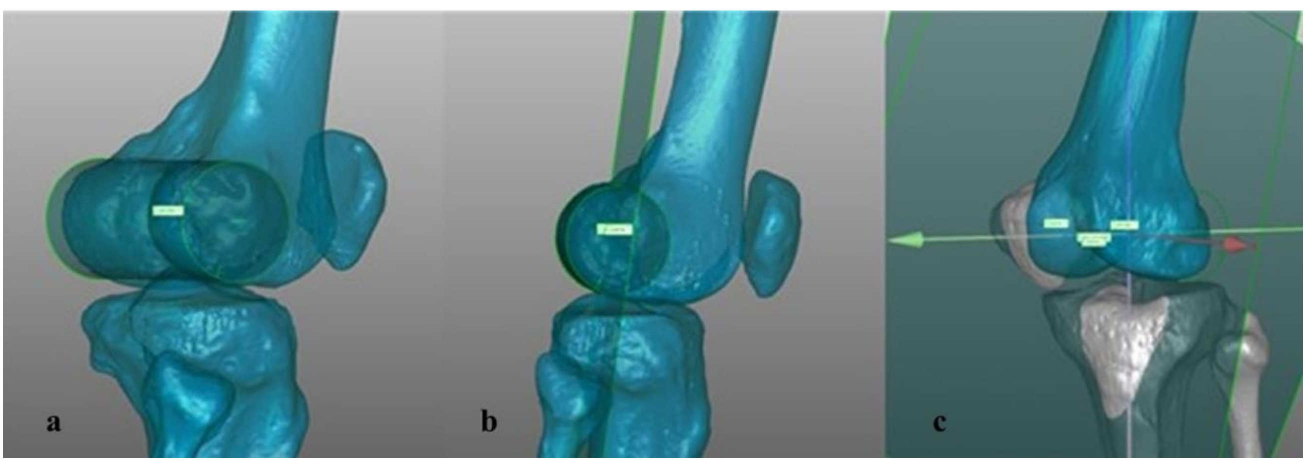

Fig. 1 Definition of coordinate system—3D model of the right knee joint; left **a**: implementation of the *x*-axis (medial–lateral), "best fit" cylinder of the femoral epicondyles with the transepicondylar centre vector as best approximation of the knee's flexion axis [17, 22]; middle **b**: implementation of the *z*-axis (longitudinal): intersecting plane between *x*-axis and the FHC as best approximation of the MFA; right **c**: implementation of the *y*-axis (anterior–posterior): recrossing the *x*- and *z*-axes incorporating the FNP → best approximation of the centre of the knee [5, 18]

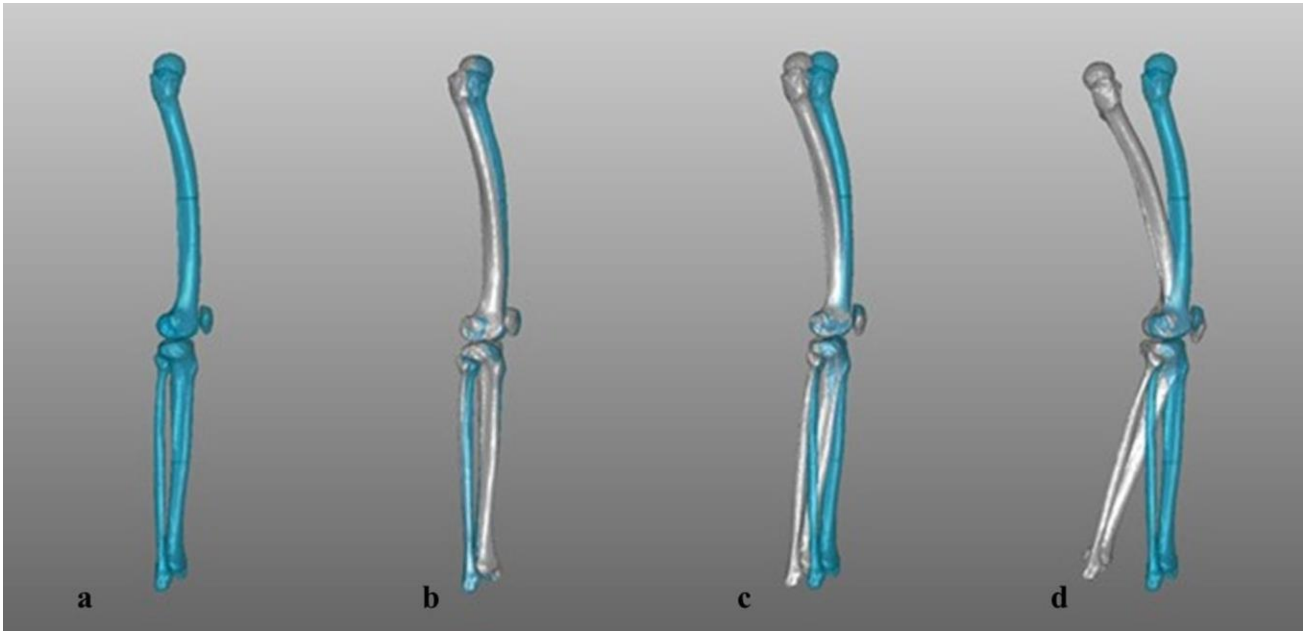

Fig. 2 Right bone model in different positions, lateral view (blue: reference zero position; white: flexed/rotated model); **a** zero position; **b** 15° external rotation; **c** 10° flexion; **d** 30° flexion with 15° internal rotation

Statistical analysis

The impact of different degrees of flexion and rotation on the measured clinical parameters MAD, HKA, mLDFA, and MPTA was analysed on a descriptive and a model-based level. Descriptive analyses focused on mean differences to the values observed without any flexion or rotation. Additionally, an individual linear mixed effects model was fitted for each of the clinical parameters (MAD, HKA, mLDFA and MPTA) using the R package lme4 [3]. With measurements given in increments of 5 and 10 degrees, respectively, rotation and flexion were treated as categorical variables with reference categories R0 and F0 for modelling purposes. In addition, a fixed effect for the leg side and a random intercept on patient level (n = 30) were included in the model. Likelihood ratio tests were applied to test for the estimated rotation and flexion effects as well as their interaction. The significance level was set to $\alpha = 0.05$ for all conducted hypothesis tests. To account for multiple testing, all p values were adjusted via the Benjamini–Hochberg method [4]. Marginal effects in terms of predicted values were visualized using the R package sjPlot [14].

All results and the related statistical calculations can also be found in the appendix, supplemental file area.

Results

All examined parameters showed highly remarkable deviations, comparing values for zero position and positions with flexion and rotation of the bone models (Figs. 3 and 4). No significant effect was found for most parameters with either rotation or flexion alone, but a significantly increasing effect in combination ($p < 0.05$). Estimated plots of the deviation to zero position for every examined combination of rotation and flexion are shown in the appendix (Figs. 5, 6, 7, 8).

Consistently high values for the conditional determination coefficient R^2, which describes the proportion of the variance explained by the combination of fixed effects (rotation and flexion) and random effects (patient effects) indicated that the established linear regression model was a very good approximation to the actual measured values from the simulation [20].

In the zero-position mean value for HKA angle was 180.1° (SD: ± 3.1°). The linear regression model (R^2 conditional = 0.93) calculated approximately a 0.03° change of measured HKA per degree limb rotation with extended knee. When the knee was 30° flexed, the change per degree increased up to 0.6° (Fig. 4).

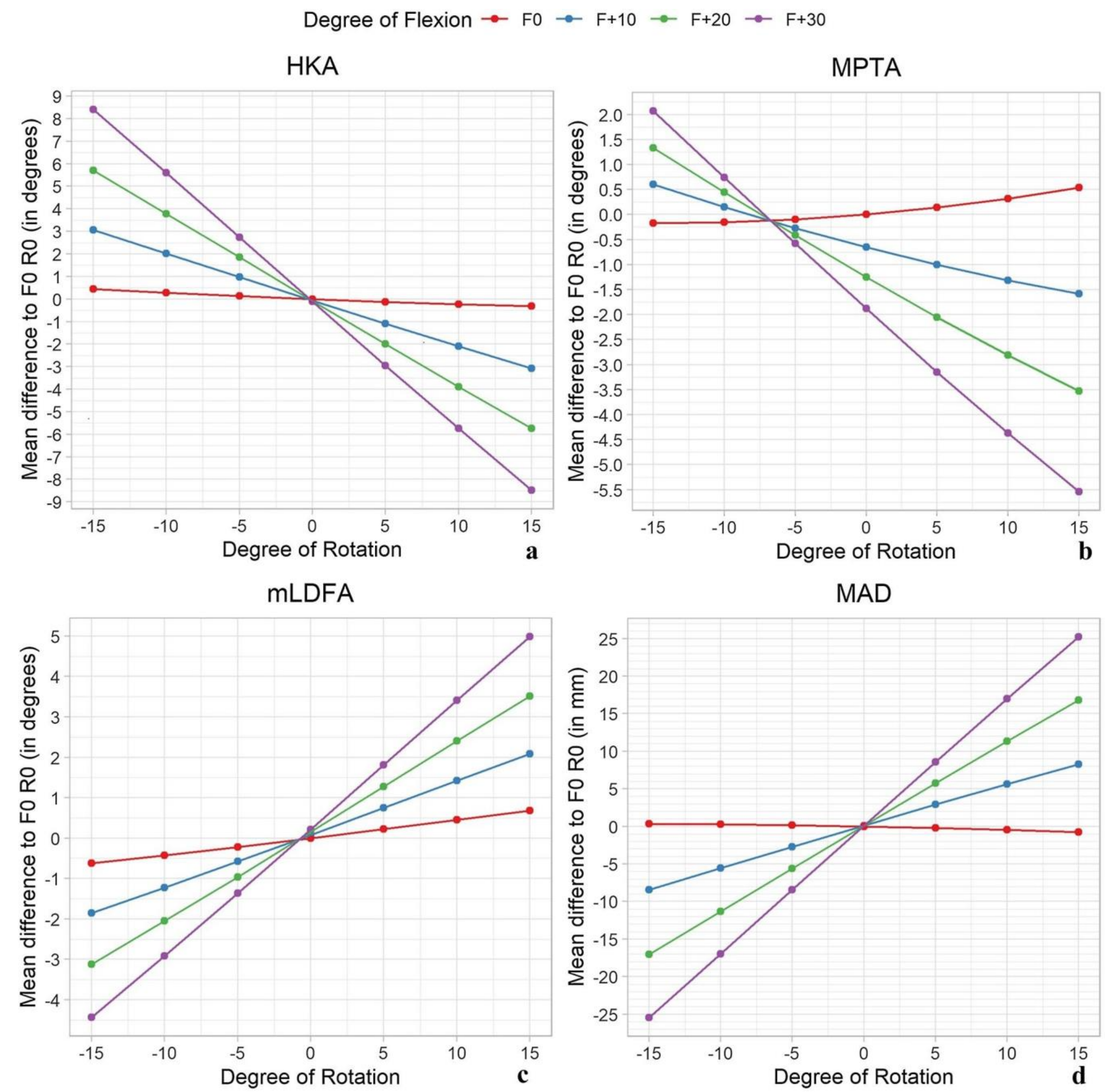

Fig. 3 Mean differences to the zero-position dependent on rotation and flexion effects measured in the simulation for HKA angle (**a**), MPTA (**b**), mLDFA (**c**) and MAD (**d**); Negative values caused by internal rotation and positive values by external rotation. Coloured graphs represent different states of flexion; *x*-axis different states of rotation; *MPTA* medial proximal tibial angle, *HKA* hip–knee–ankle angle, *MAD* mechanical axis deviation, *mLDFA* mechanical lateral distal femoral angle

The MPTA with measured mean zero position of 87.7° (SD: ± 2.6°) showed different tendencies to alter during simulation. With the knee extended, the MPTA decreased with internal rotation and increased with external rotation (Figs. 3 and 4). The linear regression model (R^2 conditional = 0.85) calculated a 0.02° change of MPTA per degree limb rotation. Additionally, the MPTA angle was most affected by flex- ion alone, compared with all other angles. A flexion of 10 degrees led to a decrease of the angle by 0.7° and 30° flexion by 1.9°. Interestingly, external rotation in combination with flexion had a higher impact on the differences to the neutral position than with internal rotation.

For the mLDFA a mean zero position of 87.2° (SD: ± 2.2°) was measured. With internal rotation the angle decreased and increased with external rotation. The linear regression model (R^2 conditional = 0.89) calculated a 0.04°

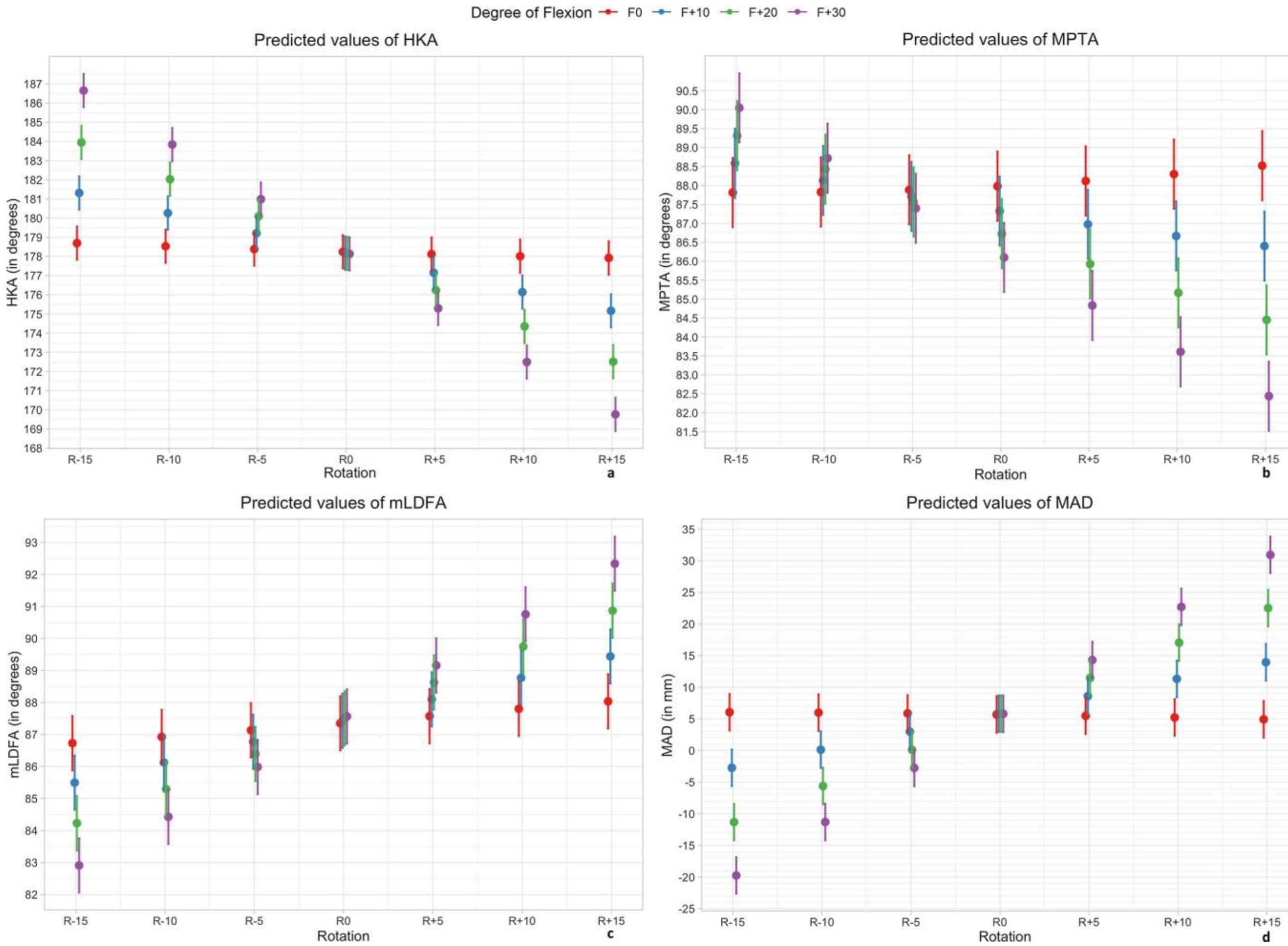

Fig. 4 Predicted values (with CI 95%) of the HKA angle (**a**), MPTA (**b**), mLDFA (**c**) and MAD (**d**); Rotation and flexion effects based on linear mixed model calculation; Negative rotation values representing internal rotation, positive external rotation; *MPTA* medial proximal tibial angle, *HKA* hip–knee–ankle angle, *MAD* mechanical axis deviation, *mLDFA* mechanical lateral distal femoral angle

change of measured mLDFA per degree limb rotation when the knee was extended. For flexion 30° the linear regression model calculated a change of 0.3° per degree additional limb rotation (Fig. 3).

The mean MAD was measured at 6.2 mm (SD: ± 8.4 mm), which is in the range Paley et al. reported as a physiological norm value 8 mm ± 7 mm [21]. Among all studied angles, the MAD was the parameter with the highest difference between singular effects and combined effects. As it is shown in Fig. 3, the combination of both led to estimated variations of approximately 25 mm in each direction.

Discussion

The most important finding of the study, was the confirmation, that rotation or flexion alone have little effect on limb alignment parameters, but when combined, these effects can reach clinically relevant values very quickly.

The demonstrated results provide a useful tool for clinicians to estimate the change in lower limb alignment parameters when radiographs are affected by extension deficits or malrotation.

In the following, our findings were compared to several studies that investigated similar questions regarding the effect of rotation on limb alignment and used comparable methods.

Lonner et al. used a singular sawbone model of a well-aligned TKA and quantified the effect of lower limb rotation and 10° flexion on the anatomic alignment [13]. They saw significant changes in tibial alignment through the additional effect of flexion, with an overall total variation in HKA of 8° from 20° internal to 20° external rotation ($p < 0.05$). In line with these findings, our values for HKA changed within a comparable range.

Similar, Kannan et al. solely investigated the influence of external rotation with additional flexion on the HKA. They addressed a similar question and concluded that flexion and rotation alone influenced the HKA < 1°, but a combination of both altered it substantially [11].

Jud et al. questioned if constitutional varus or valgus alignment (± 9°) influences the effect of flexion and rotation on alignment parameter relevantly. After performing rotation and flexion on virtual 3D models in incremental steps up to 30°, there were no relevant interpatient differences in changes of the HKA [10]. In contrast to this study, limb alignment parameters and joint angles of most of the patients investigated in our study were within the standard range (Table 1). Thus, it can be concluded that our results are probably applicable to more severe deformities.

Following Radtke et al. and several other studies, 5° steps of incremental rotation up to 15° maximum were chosen to obtain comparable results [11, 13, 23]. The linear regression model calculated a 0.05° change of the MPTA per 1° limb rotation [23]. The trend towards varus/valgus by rotation was the same in our results, but the effect was slightly smaller with 0.02° change per 1° of rotation angle in full extension. Jamali et al. predicted a significant effect on all parameters except mLDFA and anatomic lateral distal femoral angle (aLDFA) [8]. In contrast to this, no significant changes by rotation of only 3° in full extension were found. Different flexion angles seemed to be mainly responsible for the different results in the neutral origin position compared to the study of Jamali et al. thus amplifying the effects of rotation. Compared to other simulation methods, such as sawbone models, cadaveric or in vivo studies, there are some limitations regarding an appropriate biomechanical simulation. Normally LLRs are taken in weight bearing upright position with the patella pointing forward, whereas the used CT data were acquisitioned in supine position [6, 26]. Anyway, CT imaging uses a linear radiation source, whereas the X-ray beam is divergent. Therefore, CT images do not exhibit typical distortions compared to X-ray images. However, newer imaging methods such as EOS or DVT also use linear radiation [25].

Complex underlying bone deformities may significantly alter measurements, wherefore our findings are only valid for patients without severe deformities in the coronal plane. The degree of final external rotation of the tibia was set to 5° to postulate a screw home motion [12].

With these results, underlying flexion and rotation effects for patients without severe deformities can be approximated and values for calculating alignment parameters in neutral position are provided. As LLRs can only estimate rotation based on patella position or fibula overlap, while information on flexion is missing, EOS or DVT can provide coronal LLR along with sagittal and axial information that will allow the demonstrated results to be implemented in future studies and clinical practice. In addition, this study underlines the relevance of 3D imaging and 3D preoperative planning, especially when standardised positioning for LLR is not possible.

Conclusion

Axial rotation and flexion of the 3D lower limb have a huge impact on the projected 2D alignment measurements in the coronal plane. The observed effects were small for isolated rotation or flexion, but became pronounced and clinically relevant when there was a combination of both. This must be considered when evaluating X-ray images. Extension deficits of the knee make LLR prone to error and this calls into question direct postoperative alignment controls.

Supplementary Information The online version contains supplementtary material available at https://doi.org/10.1007/s00167-022-07302-x.

Author contributions JB and MJ were responsible for conceptualization, methodology, investigation, software, formal analysis, data curation, visualization, funding acquisition and writing the original draft; MW and HK for software, formal analysis, data curation, validation and review and editing. JF was responsible for conceptualization, methodology, investigation, formal analysis, data curation, visualization, writing the original draft, supervision, funding acquisition, resources and project administration.

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Data availability All data generated or analysed during this study are included in this published article (and its supplementary information files).

Declarations

Conflict of interest The authors received no financial or material sup- port for the research, authorship, and/or publication of this article.

Ethical approval The study was part of the research approved by the Ethics Committee of the Ludwig-Maximilians-University of Munich (No. 17-044).

Informed consent All authors have read and approved the manuscript.

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9. Paper II

Linear correlation between patellar positioning and rotation of the lower limb in radiographic imaging: a 3D simulation study

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Linear correlation between patellar positioning and rotation of the lower limb in radiographic imaging: a 3D simulation study

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Abstract

Purpose The purpose of this study was to quantify changes in rotation of the lower limb between image pairs based on patellar position. Additionally, we investigated the differences in alignment between centralized patellar and orthograde-positioned condyles.

Methods Three-dimensional models of 30 paired legs were aligned in neutral position with condyles orthogonal to the sagit- tal axis and then rotated internally and externally in 1° increments up to 15°. For each rotation, the deviation of the patella and the subsequent changes in alignment parameters were calculated and plotted using a linear regression model. Differences between neutral position and patellar centralization were analysed qualitatively.

Results A linear relationship between lower limb rotation and patellar position can be postulated. The regression model ($R^2 = 0.99$) calculated a change of the patellar position of − 0.9 mm per degree rotation and alignment parameters showed small changes due to rotation. The physiological lateralization of the patella at neutral position was on average − 8.3 mm (SD: ± 5.4 mm). From neutral position, internal rotation that led to a centralized patella was on average − 9.8° (SD: ± 5.2°). **Conclusion** The approximately linear dependence of the patellar position on rotation allows an inverse estimation of the rotation during image acquisition and its influence on the alignment parameters. As there is still no absolute consensus about lower limb positioning during image acquisition, data about the impact of a centralized patella compared to an orthograde condyle positioning on alignment parameters was provided.

Level of evidence IV.

Keywords Knee · Lower limb rotation · Patellar position · Long-standing radiographs · Centralized patella

Introduction

Long leg radiographs (LLR) remain essential for preoperative planning of open wedge tibial osteotomies (HTO) and total knee arthroplasties (TKA) as they allow standardized, simple, rapid image acquisition and highly sensitive identification of anatomical variations through reliable mechanical axis (MA) assessment [1]. Although three-dimensional (3D) imaging techniques with computed tomography (CT), magnetic resonance imaging (MRI) or digital volume tomography (DVT) are becoming increasingly important in surgical planning, they are still just performed in rare and complex cases, occasionally accompanied by patient-specific implants [3, 16].

In both deformity correction and TKA, LLRs are required not only for preoperative planning but also for postoperative examination of surgical precision, which is mandatory in some countries [6]. Unfortunately, there is no absolute consensus about the correct positioning of the lower limb during image acquisition yet. According to the initial definition of LLRs by D. Paley, LLRs should be obtained in true anterior–posterior (AP) view of the knee with the patella

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centred between the femoral condyles, which was the standard adjustment protocol for many years [17]. In most cases it is necessary to rotate the lower limb to achieve a position with centralized patella. Several newer studies showed that this rotation alters the alignment of the mechanical axis and various angles significantly. Consequently, all further examinations and planning procedures might be prone to error [20]. Alternatively, lower limbs should be orientated “knee forward”, with the femur condyles orthograde to the sagittal axis, parallel with the frontal reference plane, and tangential to the radiographic detector plane. This modality tends to be less influenced by present patellofemoral malalignment and tibial torsion [20].

However, the determination of angles in LLRs is highly sensible to rotational influence and prone to error, as several previous studies have demonstrated a significant decrease in medial proximal tibial angle (MPTA) and hip-knee joint angle (HKA) due to rotation [1, 2, 7]. In particular, deformities, osteoarthritis, and restricted mobility in general result in malrotation between 20° of external and internal rotation [9–11]. So far, this factor has not been considered while calculating the surgical precision between postoperative images and preoperative planning. There is already an approach to assess rotation based on tibiofibular overlap, but not yet on patellar position [12].

In this study, a linear correlation between the degree of rotation and the changes in patellar position was postulated and confirmed. It would therefore be possible to calculate alignment changes between image pairs. A further result was the clinically relevant impact on alignment parameters due to the change of focus in LLRs from true AP images with a centralized patella to knee forward images.

Methods

Overall, 60 3D-bone models of the lower limb that were already created from existing anonymized CT data of 30 randomly selected patients (18–50 years old) were used [4, 5]. As it was aimed to cover side differences between left and right limbs, both legs of each of the 30 patients were included in the study. All models were generated from post- mortem conducted CT-data that were already evaluated in previous research projects and showed limb alignment parameters that ranked within a range of reported physiological norm values. The study was approved by the Ethics Committee of the Ludwig-Maximilians-University Munich (Nr. 17–044). In zero position, mean value for HKA angle was 180.1° (SD: ± 3.1°) and for the MPTA angle 87.7° (SD: ± 2.6°). The mean mechanical axis deviation (MAD) was measured to be 6.2 mm (SD: ± 8.4 mm), which is also within the range reported by Paley et al. as the physiological norm (10 mm ± 7 mm) [2, 17, 18]. Exclusion criteria were advanced osteoarthritis of the hip or knee joint, radiographic evidence of previous realignment surgery, fractures, and any lower extremity joint replacement. The CT scans were performed on a GE HD750 CT (GE Healthcare, Chicago, IL, USA) with standardized CT parameters (slice thickness 1.25-mmm in bone kernel, helical acquisition, 120 kV, 0.8 s/ rotation, 0.984:1 pitch factor, Scan field of view (SFOV) large body, dose modulation AutomA 100–650 mA with Noise Index 8.84). Following standard procedures, the images were obtained from cranial to caudal with the patients’ legs fully extended. Digital 3D models of the legs were created using the software programs Mimics 14.0 (Materialize, Leuven, Belgium) and Geomagic Studio 2014 (3D Systems, Morrisville, NC, USA), validated rendering software for segmentation and computation, similar to 3D rendering procedures used in daily clinical practice [5]. As it was aimed to imitate LLR-images in-line with some standardized protocols, all models were aligned with the femoral epicondyles parallel to the radiographic imaging detector indicating the neutral position [20].

Overall, 37 validated and reproducible 3D landmarks on patella, femur, and tibia, which were defined and evaluated in a previous inter- and intraobserver controlled study and followed considerations of clinical practicability, were inte- grated into the models [5]. All landmarks were defined and evaluated by four senior orthopaedic surgeons that considered literature-based standard protocols for the implementation [4, 5]. An important characteristic of these landmarks is that they can be easily retrieved on conventional radiographs and enable adaptation of our 3D simulation on two-dimensional (2D) LLRs [5, 18]. As defined by Moreland et al. the centre of the femoral notch point was used as the femoral knee centre (FKC) and the midpoint of the tibial spines was used as the tibial knee centre (TKC) [5, 15]. The centre of the femoral head and the centre of the tibial articular surface of the ankle joint (AJC) defined the longitudinal axis [7, 18]. Further, the MAD was measured conventionally as the distance from the centre of the knee joint to the mechanical axis through the centre of the femoral head and AJC. The most medial and lateral points of the patella defined the patella medial pole (PMP) and the patella lateral pole (PLP), respectively [4]. Additionally, the most proximal (PRPP) and the most distal (PRDP) points on the patella ridge were marked for further considerations.

Coordinate system

The models were set in a coordinate system, which enabled us to adjust the position of every leg in an approximately identical coronal and sagittal position with the femoral epicondyles parallel to the imaging detector. Compared to this zero position, we could relate position changes due to rotation.

We first defined the medial–lateral axis by creating a geometrical "best fit" cylinder of the femoral epicondyles and using the central vector of the transepicondylar axis (Fig. 1a) [13, 19]. We hypothesized the trans-epicondylar axis to be parallel to the tangents of the femoral epicondyles and there- fore suitable for defining the coronal plane of the coordinate system, as it can be seen as an imitation of a 2D-imaging detector plane [12, 22]. The sagittal and vertical alignment was defined by two perpendicular axes orthogonal to the trans-epicondylar axis through femoral notch point and the centre of the femoral head. Thus, the zero point of the coordinate system was located at the intersection of these three axes at the centre of the epicondylar cylinder (Fig. 1b, c).

All models were aligned to the new coordinate system resulting in their individual physiological neutral position. Negative measurements indicated the lower extremity to be internally rotated and positive measurements represented external rotation around the longitudinal mechanical axis [9].

Angular measurements

The connecting lines between the centre of the femoral head and femoral notch point as well as between TKC and AJC were drawn to enable HKA angle measurement. The HKA is defined as the medial angle between those two vectors, representing the mechanical femoral axis and the mechanical tibial axis [8, 9, 18]. MAD, as the distance of the mechanical femoral axis from the centre of the knee joint, was measured by default [17, 18]. We used the most proximal lateral and medial points of the tibia to describe the medial proximal tibial angle (MPTA) [5, 18].

As we aimed to quantify the influence of lower limb rotation on patellar tracking, we defined a specific patellar position in relation to TKC and calculated the distance of both points. We imitated superimposed CT imaging by projecting the midpoint of the PLP-PMP vector and TKC onto a shared line in the same coronal and transversal plane.

Determination of the patellar position

The following formula was used to calculate absolute values for changes of the patellar alignment (Fig. 2). The calculations were performed mathematically using an application-based interface for Python scripting.

$$\text{patellar position} = \text{PLP} + (\text{PMP} - \text{PLP})/2 - [\text{TKC}]$$

PLP = Patella lateral pole (indicating the most lateral point of the patella).

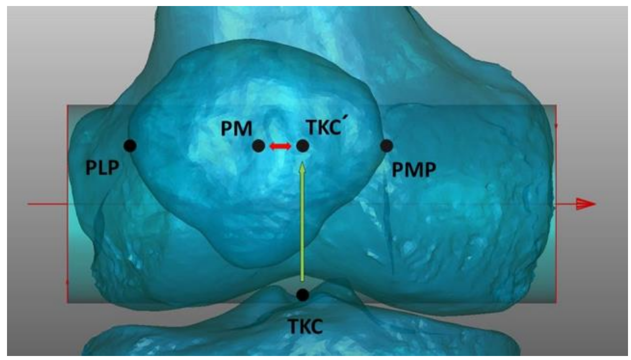

Fig. 2 Deviation of the patellar midpoint (PM) in relation to the tibial centre of the knee (TKC) in zero position; model aligned with femoral condyles parallel to the imaging detector in zero position; PM: patellar midpoint on connecting line between outer edges of the patella defined in the frontal view; green arrow shows TKC′ as a pro- jection of TKC on a shared projected line in the same transversal and coronal plane as PM; distance of projected points was measured (red arrow)

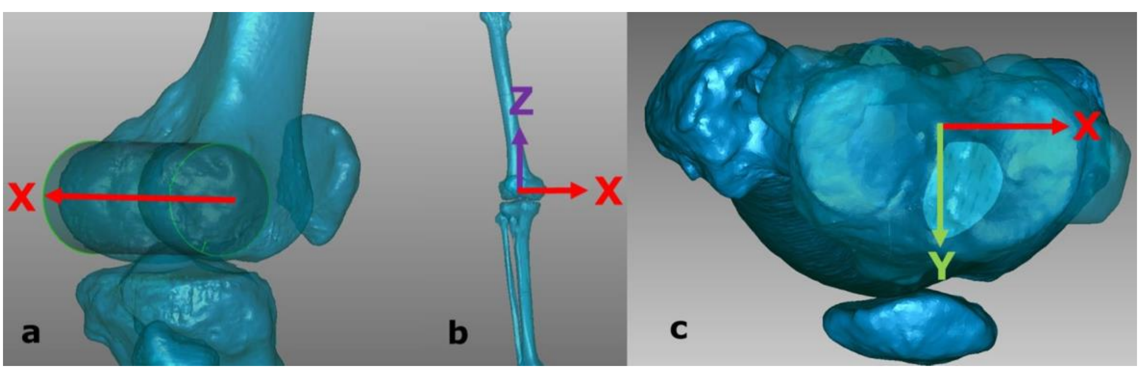

Fig. 1 Implementation of the new coordinate system; left **a** x-axis (medial–lateral, red): "best fit" cylinder of the femoral epicondyles with the trans-epicondylar-central vector as best approximation of the knee's flexion axis [13, 19]; middle **b** frontal view of the model with integrated coordinate system (x-axis = red, z-axis = violet); right **c** epicondylar view of the zero point of the coordinate system (x-axis = red, y-axis = green)

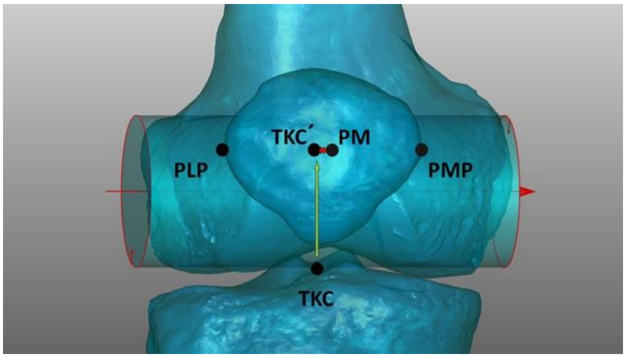

Fig. 3 Deviation of the patellar midpoint (PM) in relation to the tibial centre of the knee (TKC) with 15° internal rotation; model aligned in 15° internal rotation; patella centralized between the femoral condyles and condyles unparallel to the imaging detector; PM almost congruent with projected (green arrow) TKC′ point; changes in alignment of the patella due to rotation was measured by the length of the projected connection between PM and TKC′ (red arrow like in Fig. 2);

PMP = Patella medial pole (indicating the most medial point of the patella).

TKC = Tibial Knee Centre (Midpoint of the medial and lateral intercondylar tubercle).

3D simulation of rotation

For each model the alignment parameters in the neutral position with 0° of rotation were determined as the initial values for all subsequent measurements. The leg was then rotated along the longitudinal axis in 1° increments up to 15° internally (-) and 15° externally (+) and after every rotational step, the HKA, MAD, MPTA and the deviation of the patella from the zero position were measured (Fig. 3). Accordingly, for every model 31 positions of rotation and 1860 positions were obtained in total.

Comparison of patella centralisation with neutral position

The aim was to quantify alignment changes within image pairs of one image in true AP position with a centralized patella and one knee forward image with femoral condyles parallel to the imaging detector. For every model, the rotational position with the patella centralized between the fem- oral condyles was compared with the zero position in knee forward orientation showing the femoral condyles parallel to the imaging detector (Fig. 4).

Projection into the two-dimensional coronary plane

In addition to the common surveys of 3D angles, all meas- urements were projected into the coronal plane to mimic 2D radiographic imaging to assess the shortcomings of 2D imaging compared to 3D reality [5, 11, 14]. To gen- erate valid angular and distance measurements, all points, angles, and distances, were calculated using a Python script to ensure an automated and standardised method.

Statistical analysis

Alteration of the alignment parameters compared to the neutral position with parallel condyles and a potentially decentralized patella were analysed in a qualitative man- ner. Correlation and regression analyses were performed to examine the association between the degree of rotation and the deviation of the patella from the zero position in an exploratory way. Linear regression models were fitted to estimate expected changes in the deviation of the patella given the degree of rotation. Due to the usage of leg-specific deviations, these models implicitly take the occurrence of repeated measurements into account. Linear mixed models were used to analyse the impact of the degree of rotation on clinical parameters MAD, HKA and MPTA, with the degree of rotation here considered as categorical variable with increments of 5° (Fig. 5) [21]. All statistical analyses were

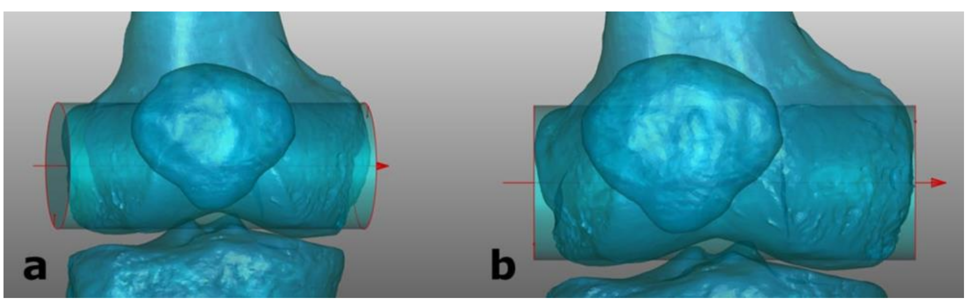

Fig. 4 Comparison between images with focus on a centralized patella and images with condyles parallel to the imaging detector; left **a** right knee in true AP position with the patella centralized between the femoral condyles and internally rotated condyles; right **b** right knee in knee forward position with femoral condyles parallel to the imaging detector and lateralized patella

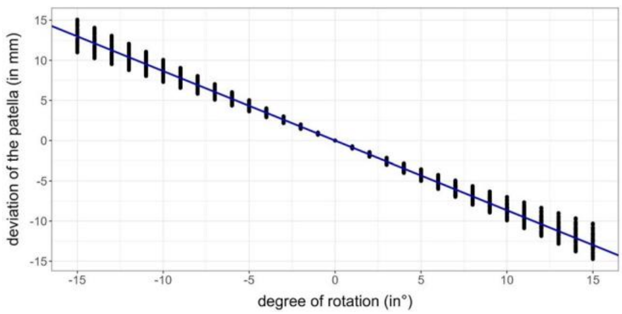

Fig. 5 Estimated deviation of the patellar position dependent on the degree of rotation; based on calculations of a linear regression model; (negative degree (in °) of rotation = internal rotation, positive degree (in °) of rotation = external rotation); (negative values of deviation (in mm) = lateralization; positive values of deviation (in mm) = medialization)

conducted using the statistical software R (R: A language and environment for statistical computing. R Foundation for Statistical Computing, Vienna, Austria; version 4.1.0).

Results

In neutral position, mean value of the patellar position was
– 8.3 mm (SD: ± 5.4 mm) externally oriented. The patella was more medialized during internal rotation and more lateralized during external rotation as it is shown in Fig. 5. Calculations of the linear regression model (R^2 = 0.99) indicated a change of patellar position by – 0.9 mm per degree limb rotation. Analysing the results of the simulation, an increase in the HKA and the MAD with internal rotation and a decrease with external rotation could be seen. Conversely, the MPTA decreased with internal rotation and increased with external rotation (Table 1). The mixed linear regression model (R^2 conditional = 0.99) calculated approximately a – 0.03° change of measured HKA per degree limb rotation and a 0.02° change of MPTA per degree limb rotation.

Most of the investigated parameters showed clinically relevant deviations due to change of orientation in model positioning from a centralized patella to parallel condyles (Table 2). Mean internal rotation that led to a centralized patella was – 9.8° (SD: ± 5.2°).

Discussion

The most important results of this study were the clinically relevant differences of alignment between true AP images with a centralized patella and knee forward images with parallel condyles. Another interesting finding was the approximately linear relationship between the degree of lower limb rotation and the patellar position due to rotation. Taken together, these results add a parameter to those currently considered when regarding the influence of rotation on lower limb alignment.

In a previous study, Maderbacher et al. already investigated malrotation in LLRs that were conducted in "true AP" view. They found large heterogeneity of rotational positions in LLRs ranging from 30° internal to 22° external rotation comparable to the values we found in our examination [11]. They further examined underlying malrotation by assessing the projection overlap of the proximal fibula and tibia using radiographic images for calculations [12]. Similar to our study, CT scans of 50 patients in different rotation positions were analysed and a strong correlation between rotation and

Table 1 Overview of deviations from zero position due to rotation for several alignment parameters derived from the (mixed) linear regression model (*n* = 60)

Deviation from zero position/ degree of rotation	– 15°	– 10°	– 5°	5°	10°	15°
Patellar position (in mm)	13	8.7	4.3	– 4.3	– 8.7	– 13
HKA (in°)	0.5	0.3	0.1	– 0.1	– 0.2	– 0.3
MAD (in mm)	0.4	0.3	0.2	– 0.2	– 0.5	– 0.8
MPTA (in°)	– 0.2	– 0.2	– 0.1	0.1	0.3	0.5

MPTA medial proximal tibial angle, *HKA* hip knee ankle angle, *MAD* mechanical axis deviation

Table 2 Alteration of alignment parameters due to change
of orientation of the model from parallel condyles to a centralized patella (*n* = 60)

Centralized patella/parallel condyles	Degree of rotation (in º)	HKA (in º)	MAD (in mm)	MPTA (in º)
Mean	– 9.8	0.2	0.7	– 0.2
SD	5.2	0.9	3.3	1.0
Max alteration internal	– 24.0	– 1.8	– 7.7	– 2.6
Max alteration external	1.0	3.1	9.1	2.4

HKA hip knee ankle angle, *MAD* mechanical axis deviation, *MPTA* medial proximal tibial angle tibiofibular overlap was found between 20° internal and 40° external rotation. A formula for determining knee rotation in radiographs was obtained by multiregression analysis and further studies confirmed their observation [11, 12]. As we wanted to establish an easier approach to predict present knee rotation and subsequent influence on alignment parameters, we focused on relative patellar position and differences between parallel femoral condyles and centralized patella.

Lonner et al. demonstrated that 5.7° valgus at 20° of internal rotation could decrease to 2.6° at 20° of external rotation, showing considerable differences to the range Maderbacher et al. take to be the common malrotation present in LLRs [10, 11].

In their CT-based 3D simulation study, Jamali et al. also investigated the influence of rotation on alignment parameters and reported values of 5.43° to 5.08° AMA between 12° internal and 12° external rotation with an average change of 0.0146° per degree of rotation [7]. The changes of HKA with 0.03° and MPTA with 0.02° per degree of rotation were in a similar range.

This study is limited in several ways. First, the investigations were performed on healthy extended legs of a random patient cohort. Possible population-dependent factors such as weight, height, or gender could not be analysed. Knee flexion, which can occur after surgery, was also not examined. Second, patients with obvious osteoarthritis or previously known deformities of the lower extremity were excluded, even though it can be assumed that the observed effects are even stronger in this group of patients. Third, possible soft tissue or ligament structures bias the position of the patella and were not considered. Fourth, parallel X-rays were assumed, like EOS imaging or DVT, but in conventional radiographic imaging the X-ray beam is divergent. Fifth, image acquisition was done in prone position in contrast to LLRs in standing weight bearing position.

The observed combination of data provides a useful tool for clinicians to predict underlying malrotation (− 0.9 mm = 1°), when image pairs show differences in patellar position. Absent intervention on the patella explaining positional changes, orientation of the femoral condyles in reference to the imaging detector should be controlled to exclude as cause for an altered image acquisition position. As there is still no absolute consensus about the optimal positioning of the limb during image acquisition, we advise the clinician to be aware of possible alignment changes that come along with the necessary rotation of the leg to obtain images with a centralized patella.

These results allow for the easy calculation of rotationally induced changes to imaging, which are minor with the knee extended and in the absence of any relevant deformity.

Conclusion

The approximately linear dependence of the patellar position on rotation allows an inverse estimation of the rotation during image acquisition and its influence on the alignment parameters. As there is still no absolute consensus about lower limb positioning during image acquisition, data on the impact that a centralized patella has on alignment parameters compared to that of an orthograde condyle positioning was provided in this study.

Author contributions MJ and JB were responsible for methodology, investigation, formal analysis, data curation, writing the original draft and visualization, MW for statistical analysis. MB, EB, WB were responsible for term, review and editing. AP and DE were responsible for supervision. JF was responsible for term, conceptualization, methodology, formal analysis, investigation, data curation, visualization, review and editing, supervision and project administration.

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Data availability The data that support the findings of this study are available from the corresponding author, MJ, upon reasonable request.

Declarations

Conflict of interest The authors have no conflicts of interest to declare.

Ethical approval The study was approved by the Ethics Committee of the Ludwig-Maximilians-University -Munich (Nr. 17-044).

Informed consent All authors have read and approved the manuscript.

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11. Appendix

The following appendix contains prizes and awards for publications that are part of this cumulative dissertation.

Furthermore, the third paper, which was mentioned under 4.2.1 in the publication list and is currently under revision in the Journal of Orthopedic Research, is presented in this appendix. Also a link to the Mendeley Database with raw data concerning this publication, is provided

https://data.mendeley.com/datasets/sy5sxn5svj/1

Open wedge high tibial osteotomy alters patellofemoral joint kinematics: a multibody simulation study

Running Title: HTO alters knee joint kinematics

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Abstract

Changes in lower limb alignment after open-wedge hight tibial osteotomy (owHTO) influence joint kinematics. The aim of this study was to investigate the morphological and kinematic changes of the knee joint, in particular the patellofemoral joint, using a multibody simulation model. OwHTO with an open tibial wedge of 6 mm to 12 mm (1 mm intervals) was virtually performed on each of 13 three-dimensional (3D) computer-aided-design-models (CAD models) derived from computer tomography scans of full-leg cadaver specimens. For each owHTO an individual biomechanical simulation model was built and knee flexion from 5° to 100° was simulated using a multibody simulation model of the native knee. Morphologic and alignment parameters as well as tibiofemoral and patellofemoral kinematic parameters were evaluated. Almost linear changes in TT-TG (0.42 mm / 1 mm wedge height) were observed which led to pathological values (TT-TG >20 mm) in 3 of 13 knees. Furthermore, 6 mm increase in osteotomy wedge height reduced lateral patellofemoral tilt by 1.0° (range: 3.2° to -2.2°) and led to a medial patellar translation of 0.7 mm (range: 1.8 mm to -1.2 mm) on average. Additionally, valgisation led to a medial translation of the tibia and a decrease in the degree of tibial internal rotation during knee flexion of approximately 0.3° / 1 mm increase in osteotomy wedge height. The increase in TT-TG and the biomechanical effects oberserved influence patellofemoral tracking which may increase retropatellar pressure and are potential risk factors for the development of anterior knee pain.

Keywords: HTO; open wedge high tibial osteotomy; kinematics; multi-body simulation; biomechanics of the knee; patellofemoral tracking

Introduction

Medial open-wedge high tibial osteotomy (owHTO) is a widely used treatment option in young and middle-aged patients with varus malalignment and early to moderate medial tibiofemoral osteoarthritis [1-3]. When performed correctly, owHTO can shift the load on varus knees to the relatively unloaded lateral compartment, relieve pain symptoms, slow the progression of degenerative process and thus lead to a normalization of the dynamic functional parameters or delay the need for knee arthroplasty [4-6]. However, surgical outcomes heavily depend on the accuracy of preoperative planning and accuracy of intraoperative procedure. In particular overcorrection after owHTO has been attributed to influence clinical outcomes and has been associated with dysfunctional patellofemoral knee kinematics, excessive shear stress at the joint surface and further progression of degernative changes in the patellofemoral joint [7-10].

Latest technological advances in the field of three-dimensional (3D) printing and computer-aided intraoperative navigation aim to further customize the surgical procedure based on the patient's individual bony anatomy. These technologies claim to enable higher surgical accuracy in achieving the preoperatively planned angulation of the corrective osteotomy with fewer outliers and less unwanted change in sagittal tibial slope [11-14]. However, when using these technologies, the standard targeted coronal correction angle is often based on the historically defined weight-bearing line percentage (WBL%), which may not reflect the ideal alignment in terms of joint kinematics and patellofemoral contact mechanics for the individual patient [3; 15]. Moreover, previous studies concluded that a customized extend of the correction after owHTO based on the pathology and thus a patient-specific alteration of targeted WBL% leads to a favorable clinical outcome and to a significant reduction in pain [16].

In contrast to coronal changes of the alignment, morphological patellofemoral changes have rarely been studied. Recently, a linear relationship in biplanar owHTO and tibial tuberosity trochlea groove distance (TT-TG) has been described, but its biomechanical implications have not yet been investigated [17].

Understanding the relationship between a change in morphological parameters and knee kinematics is crucial for optimising alignment strategies in owHTO and to further increase joint survivorship. Therefore, the purpose of this study was to investigate the relationship between different owHTO alignment parameters and simulated kinematics of the knee joint with emphasis on patellofemoral tracking using a multibody simulation model.

Methods

For this study thirteen 3D surface models derived from computed tomography (CT) scans (GE HD750 CT (GE Healthcare, Chicago, IL, USA) 1.25-mm slice thickness and intervals) of full-leg cadaver specimens virtually underwent owHTO with an open tibial wedge of 6 mm to 12mm at 1 mm intervals. A detailed description of the segmentation process, the anatomical landmarks used to define the geometric axis and performing the virtual bone cuts is provided in previous publications [18; 19]. In summary, after segmentation of the CTs using Mimics 14.0 (Materialize, Leuven, Belgium), the surface models of the legs were aligned in the coordinate system of Geomagic Studio 2014 (3D Systems, Morrisville, NC, USA) and the medial osteotomy was oriented parallel to the medial tibial slope. The biplanar cut behind the tibial tuberosity was aligned cranially [18; 19]. The patella was moved according to the translation of the tibial tuberosity, in lateral and distal direction. The virtually performed owHTO of thirteen 3D surface models with an open tibial wedge of 6 mm to 12 mm at increments of 1 mm resulted in 104 bone models that were implemented in the individual biomechanical simulation model. For each kinematic parameter assessed, the effect per degree of flexion was calculated, yielding 5120 data points each over the course of knee flexion from 5° to 100°.

Ethical approval for the use of the cadaver models was obtained from the institutional review board (No. 17-044) and the study was carried out in accordance with relevant guidelines and regulations.

Based on available landmarks, intra- and interbone morphological parameters of the knee were analysed pre- and postoperatively. Considered parameters were the TT-TG distance, the Insall-Salvati Index, medial proximal tibial angle (MPTA), hip-knee-ankle angle (HKA) as well as the medial and lateral tibial slope.

A multibody simulation model of the native knee was derived based on an existing validated model of total knee arthroplasty (TKA) built in the AnyBody Modeling System^{TM} (Anybody Technology A/S, Aalborg, Denmark) [20]. The model comprises the femoral, tibial and patellar segment, the main extensor (musculus quadriceps femoris) and flexor muscles of the knee (musculus biceps femoris & musculus semimembranosus), ligamentous structures including the collateral and cruciate ligaments and the bony surface models. External forces and muscle forces were defined as previously presented by Asseln et al. and are briefly described in the following [20]. To represent the bodyweight acting on the lower extremity, the respective external force was estimated based on the formula of Ruff et al. with respective measurements of the femoral head size, and applied at the hip joint centre [21]. Both the extensor and flexor muscles were modelled with isometric muscle force, with the cross-sectional area derived from the TLEM cadaver information and the PSCA-factor defined according to Klein Horsman [22]. For muscle recruitment a polynomial criterion of the third degree was selected [23-25]. For each patient and each owHTO version an individual biomechanical simulation model, based on the surface models provided, was built. Knee flexion from 5° to 100° was simulated in each model (figure 1). A total of five tibiofemoral and five patellofemoral kinematic parameters were evaluated for each owHTO-model and deviations to the preoperative model were calculated. For tibiofemoral kinematics, adduction-abduction rotation, internal-external rotation, medial-lateral translation, anterior-posterior translation, and proximal-distal translation were assessed. For patellofemoral kinematics, medial-lateral tilt, internal-external rotation, medial-lateral patella translation, anterior-posterior translation, and proximal-distal translation were investigated.

Differences in the mean values were analysed to identify which parameters are most affected by owHTO. We compared mean values at 100° of knee flexion, for all wedge heights and kinematics analysed.

The Kolmogorov-Smirnov test was used for evaluation of normal distribution. Statistical differences in mean values were compared using *Mann-Whitney U-test*. *Pearson's r* (correlation coefficient) was determined to measure the strength of relationship between variables.

Results

Changes in morphological parameters were observed for TT-TG, HKA, MPTA, and the lateral tibial slope with linear relationships. Correlation coefficients were high, with absolute values for the TT-TG ranging from 0.998 to 1.0 ($p<0.001$). Similar high correlations were seen in a previous study evaluating the impact of HTO solely on the TT-TG (0.99, $p<0.001$) [15]. With an increase in the osteotomy gap by 6mm, the TT-TG distance increased by 2.1 mm (±0.4) (range: 1.7 mm to 2.9 mm) on average. For three of the 13 knees, several of the HTO wedge heights led to pathological TT-TG values (>20 mm). One knee showed a pathological TT-TG of 21.6 mm already before osteotomy simulation, which increased linearly up to 25.6 mm for 12 mm wedge height. Lateral slope increased slightly by 0.5° (±0.4) (range: 0.0° to 0.8°) with an increase of 6 mm in osteotomy wedge height. As expected, an increase in osteotomy wedge height resulted in an increase in MPTA. The MPTA increased with an increase in tibial osteotomy wedge height from 6 to 12 mm, while the HKA increased accordingly (table 1).

An overview of the mean RMSE ranges for most affected tibiofemoral and patellofemoral kinematic parameters after owHTO wedge height is depicted in **Error! Reference source not found.** and figure 3, respectively. In this simulation owHTO showed little to no effect on tibiofemoral adduction-abduction rotation, anterior-posterior translation, and proximal-distal translation, as well as on patellar internal-external rotation, anterior-posterior translation, and proximal-distal translation. These kinematic parameters were therefore excluded from further analysis.

With an increase in osteotomy wedge height a medial translation of the tibia as well as a decrease in tibial internal rotation during knee flexion was observed (figure 2). With an increase in osteotomy height by 6 mm the tibia shifted medially by 0.6 mm (range: -1.1 mm to 0.2 mm) on average. For each 1-mm increase in osteotomy wedge height tibial internal rotation during knee flexion decreased by approximately 0.3° (range: 1,6° to 2,8°) on average (figure 2). Of the patellofemoral kinematic parameters investigated the increase in tibial osteotomy showed the greatest effect on mediolateral translation and the tilt of the patella (figure 3). With a 6-mm increase in osteotomy height, a reduction in lateral patellofemoral tilt of 1.0° on average (range: 3.2° to -2.2°) and a medial patellar translation of 0.7 mm (range: 1.8 mm to -1.2 mm) were observed on average. For each knee, high correlations were seen between wedge height and the affected kinematics, with correlation coefficients ranging from 0.92 to 1.0 ($p<0.001$).

The normal distribution hypothesis was rejected at the 5% level, hence for comparison of means the Mann-Whitney U-test was performed. At 100° of knee flexion, statistically significant differences in mean were found for tibiofemoral internal-external rotation starting from a wedge height of 8 mm, for tibiofemoral medial-lateral translation starting from a wedge height of 9 mm, for patellofemoral medial-lateral shift starting from a wedge height of 11 mm, and for patellofemoral tilt starting from a wedge height of 8 mm.

Discussion

In this study, the effect of increasing coronal correction in owHTO on morphological parameters of the knee and knee joint kinematics was investigated using a multibody simulation model. The key findings were an increase in TT-TG distance, a decrease in tibial internal rotation during flexion, and altered patellofemoral kinematics in terms of patella tilt and shift.

In a similar 3D simulation of owHTO in five lower extremities Hodel et al. investigated the influence of owHTO on TT-TG distance [17]. The authors reported an increase in TT-TG distance of approximately 0.5 mm/1° valgisation which is very similar to the increase of 0.42 mm/1 mm wedge height that was found in this study. Although the pathological threshold for TT-TG distance is still debated a TT-TG distance of >20 mm has been attributed to patella maltracking, instability, an increase in retropatellar pressure and patellofemoral pain [26-28]. Interestingly, in three of thirteen patients in this study, the increase in wedge height led to a TT-TG distance of >20 mm and in one of these the TT-TG distance exceeded 20 mm at 6 mm wedge height, which is frequently achieved in owHTO. Therefore, when considering owHTO, attention should be paid to preoperative TT-TG values to prevent treatment of one pathology from causing another.

In a comparable knee-simulation analysis Kuriyama et al. assessed knee kinematics during gait and squatting in a weight-bearing deep knee bend using a dynamic computer simulation with simulated changes in WBL% [32]. Similar to the simulation in this study the authors utilized a knee simulation model validated based on in vivo data that had previously been used to evaluate knee kinematics and kinetics in TKA patients. Among other parameters assessed, the authors observed a reduction in tibial internal rotation from WBL40% to WBL80% by 2.2°. In the present study a decrease in tibial internal rotation was observed, with a mean difference of 2.5° for the osteotomies of 6 mm vs 12 mm. Additionally, Kuriyama et al. reported an increase in lateral patella shift and tilt with increased wedge height [32]. In contrast, Gaasbeck et al. found decreased lateral patella shift and tilt with increased wedge height in an in vitro study [33]. One central difference between the studies lies in the consideration of changes in patellar height.

Both in our study and the one by Gaasbeck et al., changes in patellar height were found/ modelled. In contrast, Kuriyama et al. did not model changes in patellar height. Our kinematic results are in agreement with those of Gaasbeck et al., with a decrease in lateral patellar translation and shift. A potential explanation can be found in the trochlear guidance. With decreased patellar height, a stronger bony guidance through the trochlea is exerted at earlier flexion angles. Thereby, the patella may be guided more medially, despite a larger lateral component of the quadriceps muscle force. In combination with this increased lateral force component, an increase in patellofemoral pressure at the lateral facet is expected. A proposed and widely accepted mechanism for the development of anterior knee pain and the progression of patellofemoral joint osteoarthritis is an elevated stress level due to abnormal patellar lateralisation [34-36].

In a systematic review by Kataoka et al. the authors reported on twenty studies comprising 1173 patients that presented data on patellofemoral osteoarthritis or cartilage degeneration before and after owHTO [37]. The authors concluded based on available literature, that there is a tendency for patellofemoral osteoarthritis to progress after medial owHTO [37]. However, Peng et al. showed that in owHTO with coexisting medial patellofemoral osteoarthritis, patellofemoral arthroplasty achieved good clinical patellofemoral outcomes, whereas Bode et al. recommended a distal biplanar osteotomy and avoidance of overcorrection [38; 39].

The results of this study show that of all the kinematic parameters studied, an increase in the degree of coronal tibial osteotomy affects tibial internal rotation. In the coronal view of a native knee, the Q-angle describes the orientation of the quadriceps muscle force, which results from the intersection of the traction axis of the quadriceps muscle and the axis of the patellar tendon. Increasing the Q-angle has been shown to increase patellofemoral-contact pressure: shifting the patella laterally and shifting the force to the lateral patellofemoral facet [40-42]. Consequently, a decrease of tibial internal rotation with an increase of tibial wedge osteotomy as observed in this study results in an overall lateralisation of the tibial tuberosity during flexion. As a result, pressure on the lateral patellofemoral articular facet may potentially increase or cause anterior knee pain.

Overall, relevant interindividual differences in parameter changes were found during owHTO. Therefore, preoperative 3D assessment of individual effects of owHTO might be beneficial to clearly identify risk cases for e.g. pathological TT-TG distance or even to predict patient specific functional outcomes. Furthermore, patellofemoral kinematics and forces after owHTO should be investigated as a potential confounding factor for anterior knee pain and progression of patellofemoral osteoarthritis.

There are limitations in this study that need to be discussed and results interpreted accordingly. First, all CAD-models in this simulation were derived from full-leg cadaver specimens that had varying degrees of lower limb alignment with a tendency towards varus deformity and soft tissue properties were defined based on non-osteoarthritic knees. Therefore, simulating owHTO may have led to an overcorrection of coronal alignment and the kinematic simulation may not adequately reflect the biomechanical characteristics of tibial osteotomy in varus osteoarthritic knees. Although this should be seen as a major limitation, the directory of the linear data suggests similar differences in patient with varus malaligned knees. Second, the elastic properties of ligaments were modelled as either springs, e.g. cruciate and collateral ligaments, or as a rigid band, e.g. the patella tendon. Therefore, the kinematic properties during flexion may have been oversimplified because varus osteoarthritis does affect the stiffness of ligaments [43]. Future studies should incorporate pathology-adapted morphologic parameters and soft tissue property information to adequately depict the impact of owHTO on patient kinematics. Furthermore, with the model used in this study a slight external rotation of the tibia over knee flexion was found instead of the regularly observed medial pivot phenomenon. This limitation is investigated and addressed in ongoing parameter studies. However, it is not expected to alter the kinematic trends identified in this study. Finally, the simulation model has only been validated for in vitro and in vivo data of TKA. Improving multibody simulation models for owHTO may increase the understanding of the multidirectional effects of surgical intervention and thus improve the predictability of surgical outcomes. A validation of the adapted model for owHTO e.g., through fluoroscopy, would be favourable.

Conclusion

Biomechanical simulation of owHTO showed an increase in TT-TG, alterations in patellar translation and tilt, and a decrease in tibial internal rotation during flexion. This may increase retropatellar pressure and thus be a risk factor for anterior knee pain following owHTO. The effects on the patellofemoral joint, especially with a biplanar cut towards proximal, should be considered when performing owHTO. Future studies should investigate the potential of individualized planning of owHTO considering biomechanical changes to improve postoperative outcomes.

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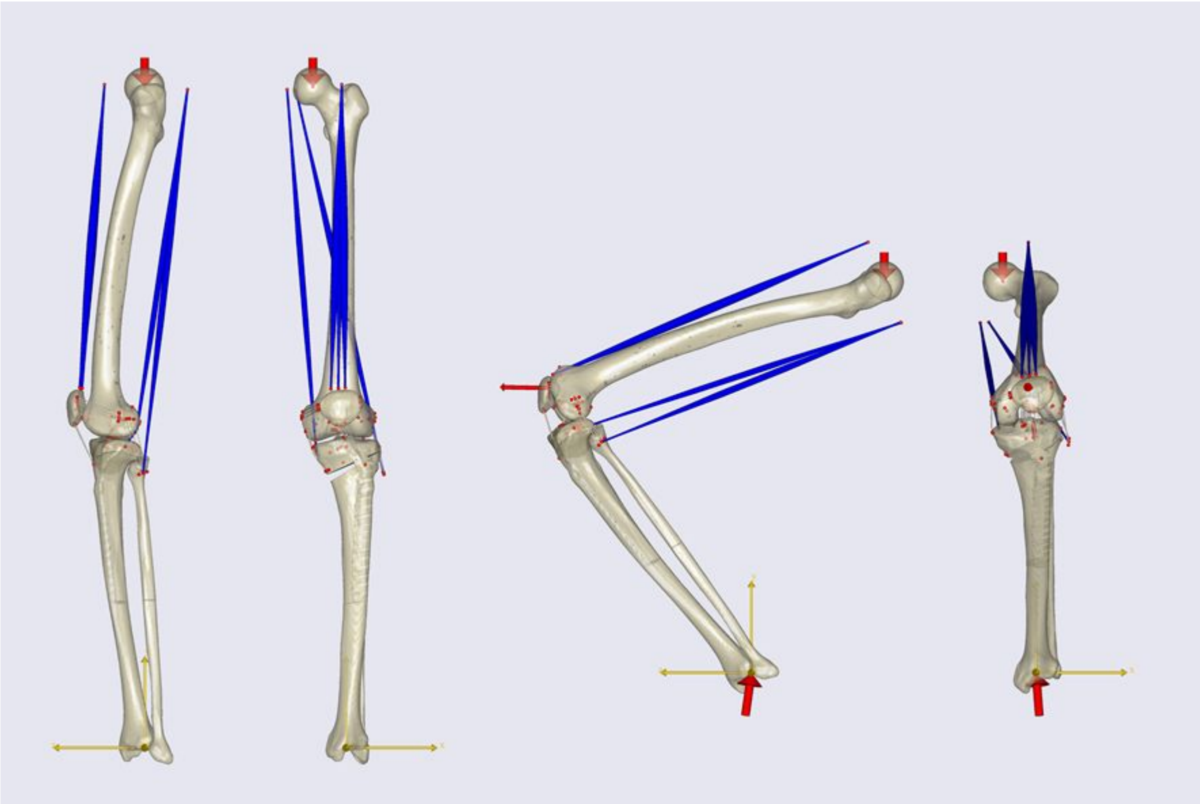
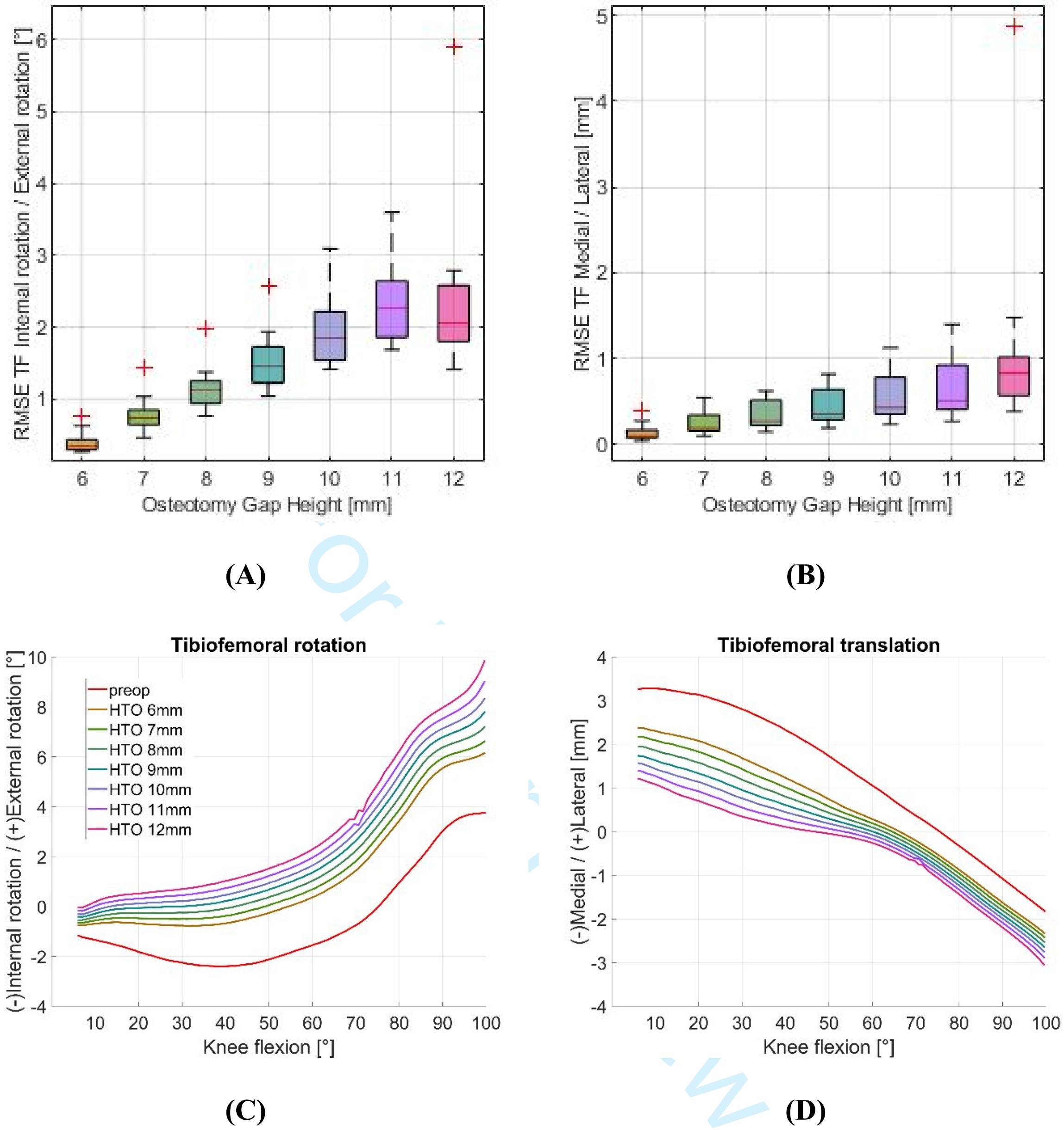
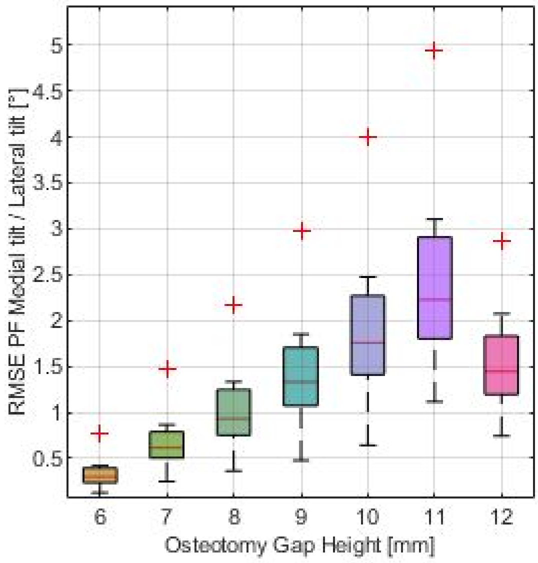


(A) (B) (C) (D)

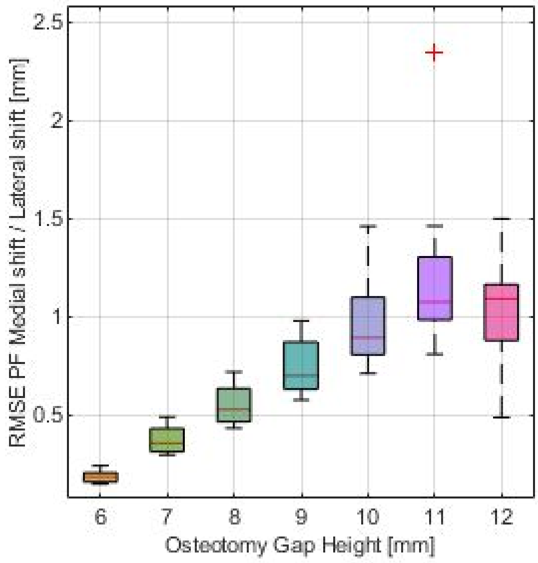
owHTO gap	TT-TG axial (mm)	Insall-Salvati Index	Medial Slope (°)	Lateral Slope (°)	MPTA (°)	HKA (°)
6mm	2.1 (±0.4)	0.0 (±0.0)	0.0 (±0.0)	0.~~4~~5 (±0.~~3~~4)	5.2 (±0.~~4~~3)	-5.2 (±0.~~4~~3)
7mm	2.~~5~~4 (±0.4)	0.0 (±0.0)	0.0 (±0.0)	0.5 (±0.4)	6.1 (±0.4)	-6.1 (±0.4)
8mm	2.~~9~~8 (±0.5)	0.1 (±0.0)	0.0 (±0.0)	0.6 (±0.~~4~~5)	7.0 (±0.~~5~~4)	-7.0 (±0.~~5~~4)
9mm	3.~~3~~2 (±0.5)	0.1 (±0.0)	0.0 (±0.0)	0.7 (±0.~~5~~6)	7.9 (±0.~~6~~5)	-7.9 (±0.~~6~~5)
10mm	3.~~8~~6 (±0.6)	0.1 (±0.0)	0.0 (±0.1)	0.~~7~~8 (±0.6)	8.~~8~~9 (±0.~~7~~5)	-8.8 (±0.~~7~~5)
11mm	4.~~2~~1 (±0.7)	0.1 (±0.0)	0.0 (±0.1)	0.~~8~~9 (±0.~~6~~7)	9.~~7~~8 (±0.~~7~~6)	-9.~~7~~8 (±0.~~7~~6)
12mm	4.~~6~~5 (±0.~~8~~7)	0.1 (±0.0)	0.1 (±0.1)	~~0.9~~1.0 (±0.~~7~~8)	10.~~6~~7 (±0.~~8~~6)	-10.~~6~~7 (±0.~~8~~6)
0mm (mean)	~~15.9~~13.6 (±~~2.4~~3.6)	0.8 (±0.1)	~~9.1~~10.2 (±~~1.7~~3.3)	~~8.6~~9.0 (±~~2.1~~3.7)	~~89.8~~88.1 (±~~1.7~~3.0)	~~180.6~~178.1 (±~~1.6~~2.7)

Table 1: Mean changes in knee morphological parameters compared to mean values before owHTO; *standard deviation in brackets*

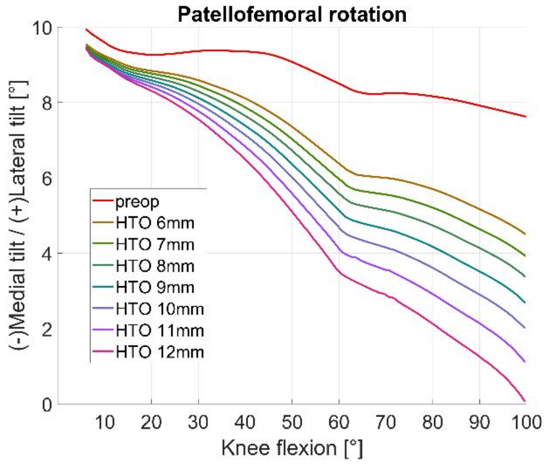

RMSE TF Internal rotation / External rotation [°]
Osteotomy Gap Height [mm]
(A)
RMSE TF Medial / Lateral [mm]
Osteotomy Gap Height [mm]
(B)
Tibiofemoral rotation
(-)Internal rotation / (+)External rotation [°]
preop
HTO 6mm
HTO 7mm
HTO 8mm
HTO 9mm
HTO 10mm
HTO 11mm
HTO 12mm
Knee flexion [°]
(C)
Tibiofemoral translation
(-)Medial / (+)Lateral [mm]
Knee flexion [°]
(D)


RMSE PF Medial tilt / Lateral tilt [°]
Osteotomy Gap Height [mm]
6
7
8
9
10
11
12

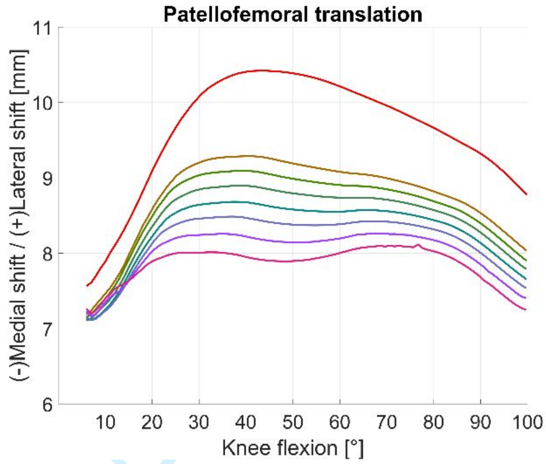
(A)


RMSE PF Medial shift / Lateral shift [mm]
Osteotomy Gap Height [mm]
6
7
8
9
10
11
12

(B)


Patellofemoral rotation
(-)Medial tilt / (+)Lateral tilt [°]
Knee flexion [°]
preop
HTO 6mm
HTO 7mm
HTO 8mm
HTO 9mm
HTO 10mm
HTO 11mm
HTO 12mm

(C)


Patellofemoral translation
(-)Medial shift / (+)Lateral shift [mm]
Knee flexion [°]

(D)

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verleiht

Herrn Dr. med. Maximilian Jörgens & Herrn cand. med. Josef Brunner

Muskuloskelettales Universitätszentrum München, Klinikum der Ludwig-Maximilians-Universität München

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Changes in lower limb alignment due to flexion and rotation - a 3D simulation of radiographic measurements

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