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A musculoskeletal simulation study on the biomechanics of the lower extremity during the golf swing

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Purpose: This study aimed to investigate the mechanisms of the human body dynamics during the golf swing through a musculoskeletal simulation for different types of subjects to better understand the biomechanical aspects of the golf sport. *Methods*: The lower extremity model of the AnyBody™ modeling system was adapted to an advanced knee model to capture the large knee axial rotations during golf swing using an inverse dynamic musculoskeletal approach. Swings performed by three golfers, one with apparent osteoarthritis in both knee joints, one with bilateral total knee arthroplasty and one healthy, were captured in a motion capture laboratory and simulated. *Results*: The golf swing generated a high axial rotation in the knee joint (approximately 25–31°). Ball impact represented the most critical time event, near which maximum loading of the lower extremity was observed, e.g., knee compression force raised to a maximum of 329% body weight. The fast rate of knee loading by large compression forces prior to ball impact was identified as a potential cause of knee injury in golfers. *Conclusions*: Our findings are comparable with previous experimental and computational studies, and the proposed musculoskeletal model can be employed to provide valuable information to clinicians and scientists, e.g., on the biomechanics of the lower extremities after total knee arthroplasty during golf swing.

Key words: *golf sport, musculoskeletal multibody model, inverse dynamics, lower limb, knee injuries*

1. Introduction

Golf is a popular sport worldwide, with approximately 37.5 and 10.6 million participants in the U.S. and Europe, respectively [9], [16]. A complex spatial sequence of whole-body motions generates considerable forces and rotational movements. Although it is commonly considered a low-impact sport [7], injuries can still occur.

In this context, a systematic review study found that up to 7% of golf-related injuries are associated with the knee joint [10]. It was reported that more than 53% of all assessed active golfers with total knee arthroplasty (TKA) experienced radiolucency [30]. Additionally, playing golf can lead to severe wear of the endoprosthesis due to increased rotational loading [51]. Therefore, it is essential to investigate the biomechanics of the knee joint in golfers in more detail to provide better insight into the mechanism of golf-related injuries and provide more evidential support for decisions regarding returning to golf or starting to play, e.g., after TKA.

To achieve this goal, multiple researchers have studied knee kinematics in different cohorts of golfers based on factors such as gender [6], club type [48], age [47] and skill level [49], and compared them to daily activities [41]. The kinetics of the golf swing, includ-

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ing knee compression force, joint moments, and ground reaction forces, have been studied using instrumented implants [14], [38] and computational analyses [6], [11], [12], [18], [41], [42]. Moreover, the activation pattern of muscles during golf has been investigated using EMG signals [8], [34].

These studies have provided valuable information on the joint biomechanics of the golf swing. However, no systematic research has employed a musculoskeletal simulation framework to examine the lower extremity muscle force profiles of average-skilled golfers with different knee conditions [7], [35]. Estimating muscle forces also plays an essential role in joint kinetics estimation [20]. To this end, the current study aimed to establish a musculoskeletal simulation framework for golf swing and examine its practicability by simulating swings of different subjects, e.g., healthy golfers and individuals with knee pathologies.

2. Materials and methods

2.1. Participants

Three right-handed male golfers participated in this study (Table 1): one patient with Osteoarthritis OA in both knees (Kellgren Lawrence 4), who is still active, another patient who underwent TKA in both knees with a right knee endoprosthesis of Enduro modular revision rotating hinge knee (Aesculap, Tuttlingen, Germany; implanted primarily in 2006) revised in 2014, and a left knee endoprosthesis of Natural-Knee® II (Zimmer Biomet, Winterthur, Switzerland, implanted primarily on 2006), and a healthy subject without any pain, indications or history of knee pathology. These three participants were average-skilled recreational golfers with a handicap index of 15–30. Before the experiments, all the subjects provided appropriate informed consent, and the study was approved by the local ethics committee of the Rostock University Medical Center (ID: A2020-0191).

Table 1. Demographic data for the participants (n = 3 subjects)

Characteristics	Mean ± SD	Range
Age [years]	64 ± 9.8	56–75
Body Height [m]	1.80 ± 0.02	1.77–1.81
Body Mass [kg]	103.93 ± 17.58	89–123
Experience [years]	>25	–

Data acquisition

To capture the motion of the subjects through video analysis, the Gait Real-Time Analysis Interactive Lab system (GRAIL, Motek Medical, Amsterdam, The Netherlands) was utilized with an integrated 10--camera motion capture system operating at 100 Hz (Vicon Bonita B10, Vicon Metrics, Ltd., Oxford, United Kingdom) and ground reaction forces (GRFs) were collected by a split-belt treadmill with two embedded force plates (Motek Force link, Houten, the Netherlands) operating at 1000 Hz. A set of 26 retroreflective skin markers was attached to each participant's body according to the Human Body Model 2.0 (HBM, Motek, Amsterdam, The Netherlands) [53]. This marker protocol is selected for the current study because it provides robustness for the musculoskeletal modeling by adding redundancy, which is of great importance for the golf swing, which includes a wide range of internal and external rotation of the knee joint [33], [46]. Redundancy means there are more markers than required to capture the motion. More information regarding marker attachment and GRAIL specifications and setup can be found in a previous study [44]. The participating golfers chose their own stance condition and no restrictions were applied to their swing. Before recording, the swing trials intended for evaluation, a warm-up including typical stretch and golf swings was performed. Within the laboratory environment, each participant conducted several swing trials for self-adaptation to the task. A static calibration in the anatomical standing condition for 0.5 seconds was performed for each participant. Following the familiarization phase, each participant was asked to perform swings and strike the ball toward a vertical target placed 2 meters away until at least five swings were obtained for analysis.

Equipment

Participants performed swing trials while standing on two force plates located side by side. A golf launch pad (SKLZ, North Carolina, USA) was laid on the force plates in front of subjects' feet. An indoor training softball with a diameter of 42 mm was used. All participants throughout the study used a unique nine-iron club fitted with a regular shaft. Golfers chose the sports shoes and shorts of their choice.

Data processing

Raw marker trajectories and ground reaction forces were exported from VICON Nexus 2.9 (Vicon Motion Systems, Ltd, Oxford, UK). Marker trajectories were smoothed using a second-order zero-phase low-pass

Butterworth filter with a cut-off frequency of 5 Hz adopted from [42], while ground reaction forces were smoothed using a fourth-order Butterworth low-pass digital filter with a cut-off frequency of 10 Hz [18].

Swing phases and events

For a systematic biomechanical analysis, the golf swing is divided into separated sequential phases and events (Fig. 1) [23]. Each swing starts with the ball address event followed by the top of the back swing (TBS), trail horizontal club, ball impact (BI), where the golfer hits the ball, lead horizontal club, and the end the of the swing. Between each of the two successor events, a phase is defined. The first phase is back swing, which occurs between the ball address and TBS; the second is down swing (DS), which is divided into the forward swing (FS) and acceleration (ACC); the third is follow through, which starts by early follow through (EFT) associated with body deceleration and ends in the late follow through (LFT). The moment the club head markers first reached their initial mediolateral position at the beginning of the takeaway phases was characterized as BI event [32]. The lead limb refers to the limb positioned closer to the target; e.g., for a right--handed person the left side is the lead side of the body (Fig. 1).

Body Technology A/S, Aalborg, Denmark) [13] was used in this study as a basis to establish a musculoskeletal simulation framework. As for the kinematic chain of the human body, the hip joint is modeled as a ball-and-socket (spherical) joint. Subtalar and foot joints are modeled as hinge joints to model ankle plantar, dorsi flexion, and eversion-inversion, respectively. The original AnyBody™ knee model is defined as a hinge joint with one degree of freedom that mimics only the flexion-extension rotation of the knee. This simplification has the potential to result in kinematic and kinetic inaccuracies, which can impact the functioning of the thigh muscles [13]. In the case of the golf swing, which includes larger internal-external (axial) knee rotations [33], [46] compared to gait, it was essential to develop the knee joint model in this study as a universal joint. In this manner, both flexion-extension and internal-external rotational degrees of freedom were considered. Defined joints constrain the relative motion between eleven bone segments (pelvis, femurs, shanks, patellae, talus and feet) representing the lower extremity and a simplified torso. No degrees of freedom were considered for the patellofemoral joint. Therefore, the simulation model had a total of 23 degrees of freedom. Fifty-five muscle structures were implemented to actuate each lower extremity. Muscles with large attachment sites, as well as long

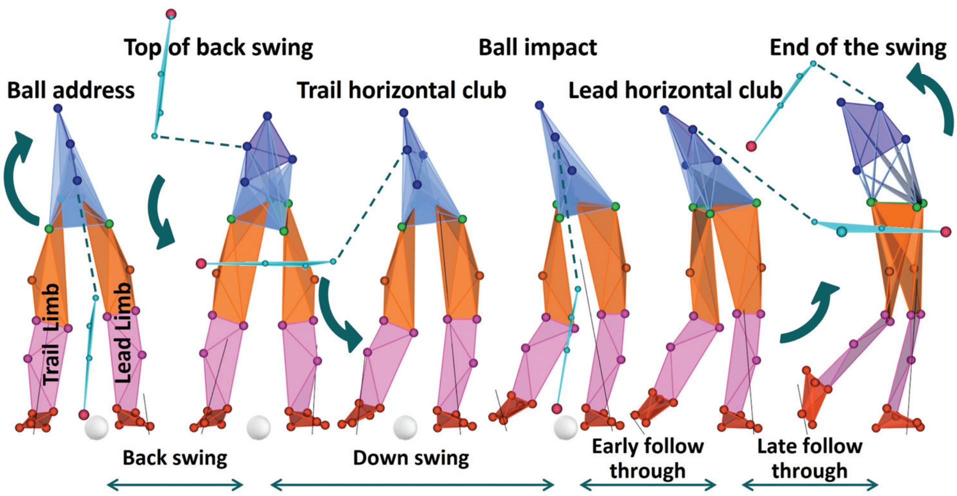

Fig. 1. This figure depicts different phases and events of a golf swing, highlighting distinct body segments with corresponding colors: thorax (blue), pelvis (green), thigh (orange), shank (purple), foot (red), and the golf club (cyan). The directional green arrows illustrate the rotation of the golf club between different swing phases. Beneath the feet, solid black lines represent the ground reaction forces, while green dashed lines symbolize an imaginary representation of the body arms for better visualization

Musculoskeletal model of the lower extremity

The Twente Lower Extremity Model version 2.0 (TLEM 2.0) included in the Mocap Lower-Body model from the AnyBody™ Modeling System 7.3.1 (Any-

muscles, were divided into smaller parallel and or serial sections. In total, more than 160 muscle elements were included in each leg in the musculoskeletal model of the lower extremity, with each muscle element modeled as a force element [13].

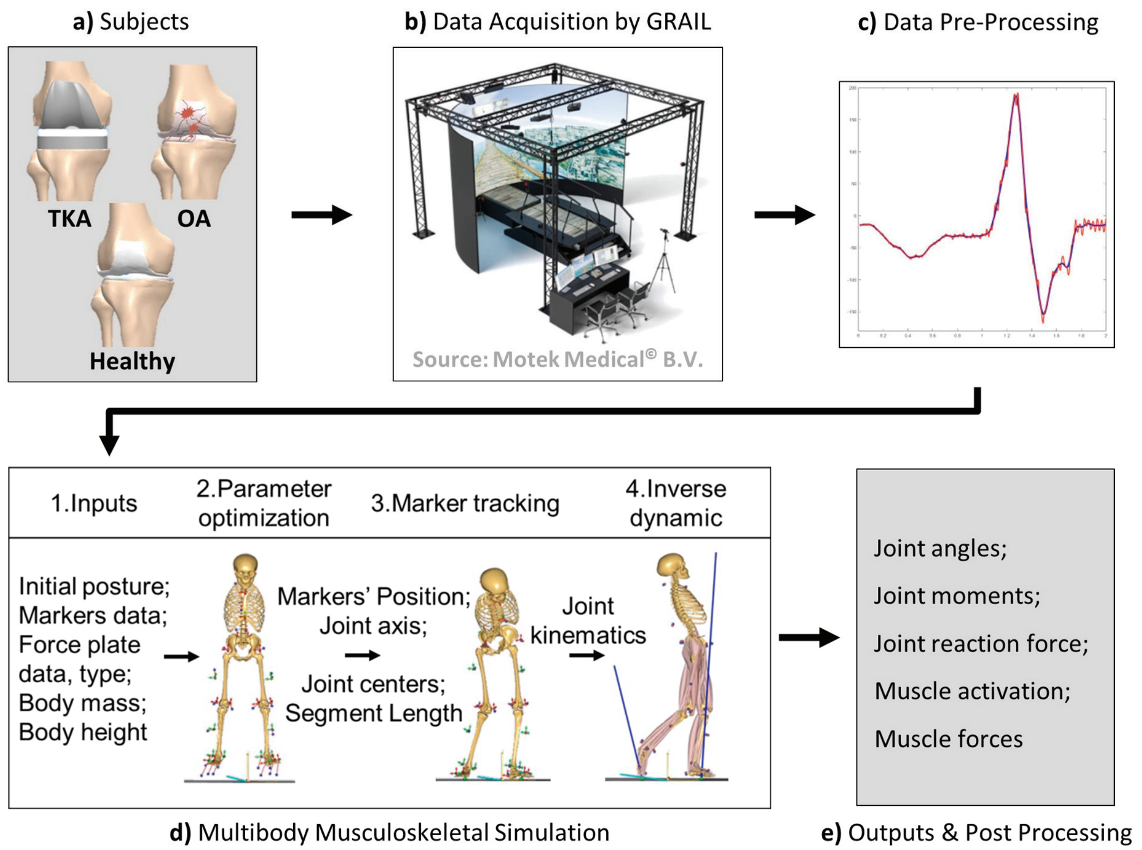

Fig. 2. Musculoskeletal simulation framework for investigating biomechanics of the lower extremity during golf swing

After data acquisition by the GRAIL (Fig. 2b) and a pre-processing stage (Fig. 2c), acquired marker trajectories, GRF and subject's body height and weight served as inputs for this musculoskeletal model (Fig. 2d.1). The generic human model was scaled through a parameter optimization to fit the measured body height and weight of each specific participant (Fig. 2d.2). Linear scaling of the length of the segments in this model (through parameter identification) was performed according to each golfer's body height, body weight, static trial, and dynamic trials data [29], which, consequently, personalize other factors such as muscle lever arms and calibrate the marker positions with respect to the human body. To scale the isometric muscle strength of each muscle and segment size (in addition to the length of the segments), the "Length-Mass-Fat" scaling law was used [43]. The marker trajectories (experimental markers) were used as input for an inverse kinematic (marker tracking) analysis to calculate the joint angles through an optimization (Fig. 2d.3). In this context, the model markers track the experimental markers by minimizing the least square error between model markers and the corresponding experimental markers [2]. Subsequently, joint angles as well as ground reaction forces were used for an inverse dynamic analysis followed by a static optimization to calculate the joint loading and muscle forces (Fig. 2d.4). More precisely, the muscle force-sharing problem was fulfilled by minimization of a 3rd order polynomial function of muscle activations. Mathematical muscle models were subsequently used to calculate the muscle forces based on muscle activation [1]. By means of this musculoskeletal model, different biomechanical factors of the golf swing, including knee joint angles and range of motion, knee joint moments, knee compression forces, and lower extremity muscle forces (Fig. 2e), were calculated in this study for the healthy and pathological golfers, as presented in Section 3. Moreover, the results were compared to the current state of research.

3. Results

Knee joint angle and range of motion

From the ball address up to near of the beginning of the down swing (Fig. 3), the flexion angle of the knee was increased gradually, followed by a rapid extension of the knee to the end of the early follow through, after which it remained almost constant. The peak flexion angle for the knee of the lead leg oc-

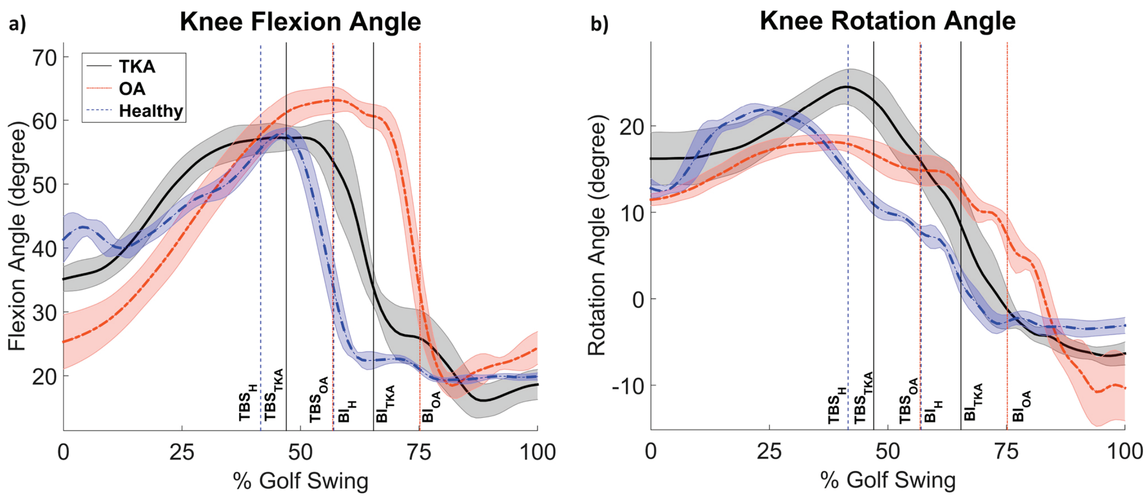

Fig. 3. Joint angles of the lead knee. (a) flexion (b) internal/external rotation (with positive values indicating internal rotation of the femur relative at the tibia) at different key events throughout the full golf swing

Table 2. Peak value of the mean knee joint angles [°] and moments [Nm/kg] in each anatomical direction for $N = 5$ trials

Direction	Peak of joint angles			Direction	Peak of joint moments		
	TKA	OA	Healthy		TKA	OA	Healthy
Flexion	57.27 ± 1.8	63.21 ± 1.9	57.89 ± 0.8	Flexion	0.65 ± 0.3	0.72 ± 0.1	0.64 ± 0.1
Internal rotation	24.58 ± 1.0	18.13 ± 0.8	21.86 ± 0.6	Extension	0.44 ± 0.1	0.34 ± 0.1	0.73 ± 0.1
External rotation	6.6 ± 1.1	10.78 ± 4.0	3.47 ± 0.9	External rotation	0.13 ± 0.0	0.14 ± 0.0	0.21 ± 0.0

curred after the top of the back swing (Table 2, Fig. 3a), and the minimum flexion angle occurred near the beginning of the late follow through. Range of knee mean flexion was 41.2, 44.7 and 38.6° for TKA, OA, and Healthy subjects, respectively.

Internal rotation of the knee was defined as the inward rotation of the lead (left) femur with respect to the tibia. Participants internally rotated the knee to a maximum value near the TBS (Table 2). Then they applied external rotation to the end of EFT, followed by an almost constant angle to the end of the swing (Fig. 3b). The range of knee mean rotation was 31.2, 28.9 and 25.3° for TKA, OA, and Healthy subjects, respectively.

Knee joint moments

Concerning the joint moments, the lead leg experienced a peak knee joint extension moment around the middle of the DS as the leg prepared for the ball impact, and the peak flexion moment happened just after

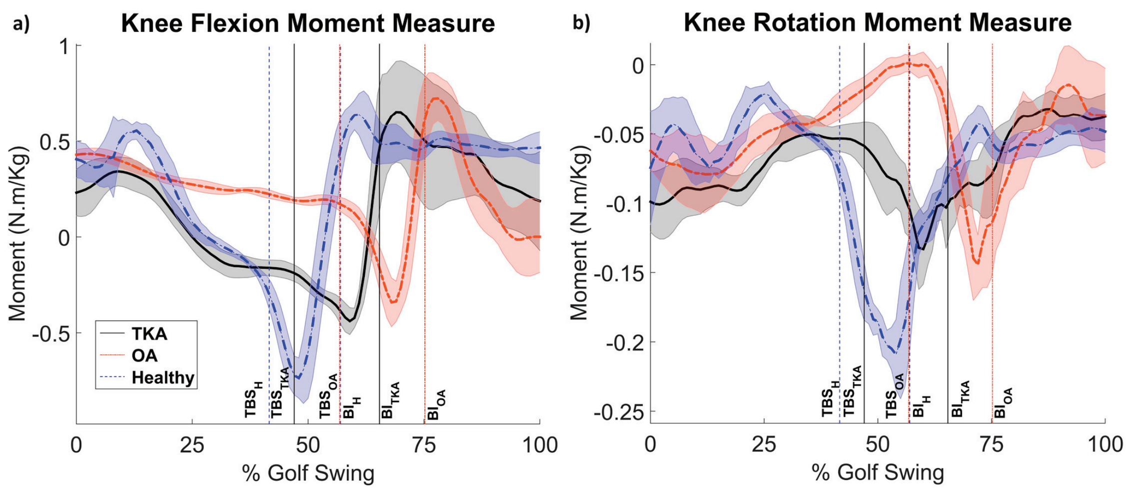

Fig. 4. Joint moments of the lead knee: (a) flexion (+)/extension (–), (b) internal rotation moments, at different key events throughout the full golf swing

the BI (Fig. 4a). Both TKA and Healthy subjects started and continued the swing by applying flexion moment to the knee. For them, the flexion moment was decreased to zero in the middle of the Back Swing and applied the extension moment to the knee at the TBS. However, the OA subject still applied flexion moment to the knee at the TBS and continued to the middle of the DS. The peak value for the knee internal rotation moment occurred in the late DS before the BI (Fig. 4b, Table 2).

For both sagittal and transverse planes, TKA and OA subjects showed almost the same peak value of mean joint moment (Table 2, Fig. 4). Although the mean range of knee motion for the Healthy subject was smaller than the other two subjects, the mean extension moment was almost doubled for this subject.

Knee compression force and vertical ground reaction forces

The peak value of the mean knee compression force occurred to the Healthy subject before the ball impact and for the TKA and OA subjects at the beginning of the early follow through after the ball impact (Fig. 5a, Table 3). This value increased drastically in the magnitude during the DS from the TBS to BI by 18.35 (242.4%), 21.30 (665.6%), and 16.14 (138%) N/kg for the TKA, OA and Healthy subjects, respectively.

Peak values for the vertical GRF are reported in Table 3. It is noteworthy that the OA subject flexed his lead knee to a maximum level at the TBS (Fig. 3a), which lowered the vertical ground reaction force to a minimum level near zero at this time event (Fig. 5b).

Muscle forces of the lower extremity

This section presents the contribution of lower extremity muscles in motion generation throughout the golf swing phases. While knee extensors (m. rectus femoris and m. vastus lateralis) showed the peak muscle force near the middle of the DS, lead knee flexors (m. semimembranosus, m. semitendinosus, and m. biceps femoris) produced the peak muscle force after the BI (Fig. 6d–f), which is in line with peak flexion and extension moments timing (Fig. 4a). Unlike the knee extensor muscles, which were not active at the beginning of the swing, knee flexor muscles generated a medium level of muscle force at this stage. Forces of knee flexor muscles were decreased to zero by the middle of the BS. At this point, extensor forces increased to a maximum level and then started decreasing, where the knee flexor muscles began growing again. This timing agrees with generated knee joint flexor and extensor moments (Fig. 4a).

The most active phase of the golf swing for the lead leg, based on the activation of examined muscles, was recognized to be EFT (Table 4, Fig. 6). The muscles of the lead leg generated much more force at BI than at TBS (Table 4).

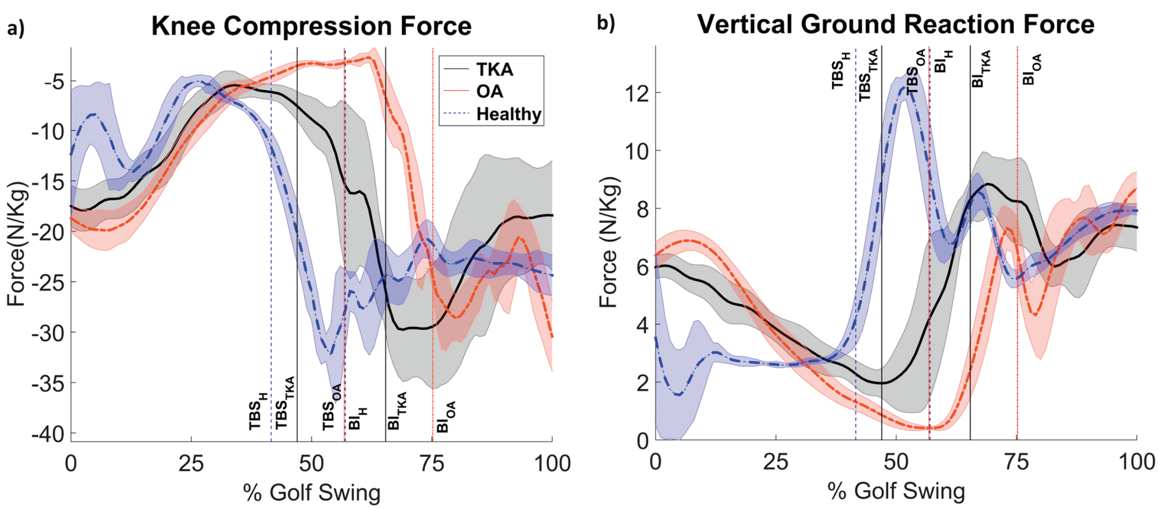

Fig. 5. (a) Lead knee compression force, (b) vertical ground reaction force applied to the lead foot at different key events throughout the full golf swing

Table 3. Peak mean knee compression and vertical ground reaction forces [N/kg] for $N = 5$ trials

	TKA	OA	Healthy
Knee compression force	29.8 ± 5.1	28.6 ± 2.2	32.3 ± 3.0
Vertical ground reaction force	8.8 ± 1.1	7.3 ± 0.7	12.2 ± 0.9

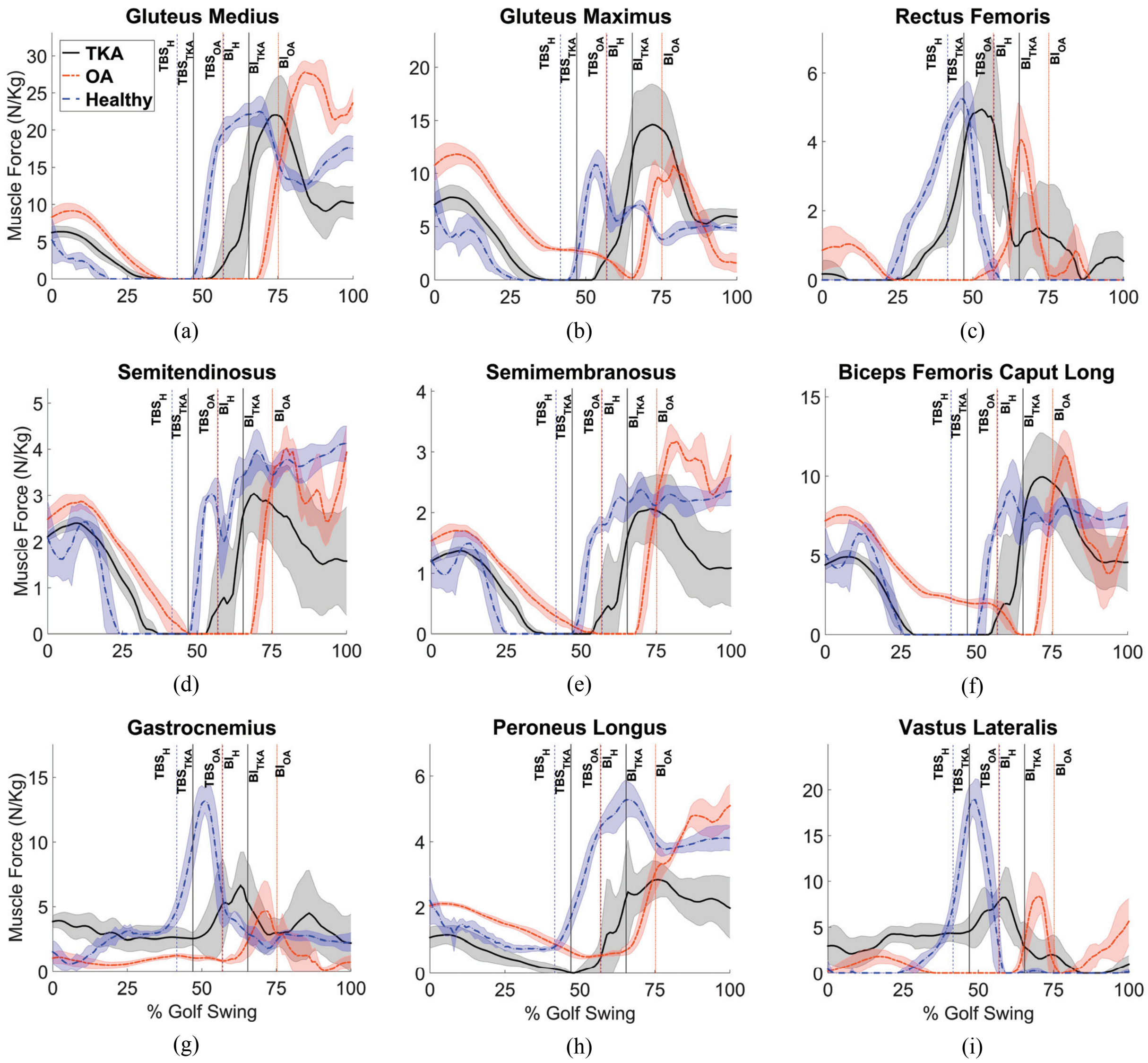

Fig. 6. Muscle forces [N/kg] in the lead lower extremity for TKA (Black), OA (Red) and Healthy (Blue) subjects at different key events throughout the full golf swing

Table 4. Mean muscle forces [N/kg] of the lead leg for TKA, OA and healthy subjects

	TKA subject				OA subject				Healthy subject			
	Peak (%GS)	TBS	BI	MAP	Peak (%GS)	TBS	BI	MAP	Peak (%GS)	TBS	BI	MAP
GMD	22.0 (74)	0	13.2	EFT	27.8 (84)	1.9	14.7	EFT	22.5 (69)	0	19.8	EFT
GMX	14.6 (72)	0	11.5	EFT	10.7 (79)	2.0	9.3	EFT	10.8 (53)	0	7.8	ACC
RF	4.9 (53)	3.8	1.0	FS	4.1 (66)	0.3	0.1	DS	5.3 (46)	4.5	0.2	FS
VL	8.2 (58)	4.5	2.9	ACC	8.3 (70)	0	1.0	DS	18.9 (48)	7.3	2.3	DS
ST	3.0 (69)	0	2.6	EFT	4.0 (80)	0	3.4	EFT	4.0 (70)	0	2.6	LFT
SM	2.1 (73)	0	1.6	LFT	3.2 (82)	0	2.2	LFT	2.4 (70)	0	1.8	LFT
BF	10.0 (72)	0	7.1	EFT	11.3 (79)	1.7	8.6	EFT	9.0 (61)	0	7.3	EFT
GN	6.7 (63)	2.5	5.3	ACC	4.6 (72)	0.8	2.8	ACC	13.2 (51)	4.8	5.6	ACC
PL	2.8 (76)	0	2.3	EFT	5.1 (100)	0.5	3.1	LFT	5.3 (66)	0.8	4.5	EFT
Sum	74.4 (68)	10.9	47.6	EFT	78.9 (79)	7.2	45.2	EFT	91.4 (59)	17.5	51.5	FT

TKA – Total knee arthroplasty, OA – Osteoarthritis, GS – Golf swing, MAP – Most active phase, GMD – gluteus medius, GM – gluteus maximus, RF – rectus femoris. ST – semitendinosus, SM – semimembranosus, BF – biceps femoris, GN – gastrocnemius, PL – proneus longus, VL – vastus lateralis, DS – down swing, FS – forward swing, ACC – acceleration, EFT – early follow through, LFT – late follow through.

4. Discussion

In this study, a musculoskeletal model is proposed to investigate the biomechanics of the lower extremity during a golf swing for different types of subjects to gain a better insight into the musculoskeletal biomechanical aspects of the golf swing. Although inverse dynamics is still the most frequent approach to estimating the human joint loadings in sports biomechanics, this approach does not consider the internal forces while estimating joint kinetics [20]. Musculoskeletal modeling, on the other hand, estimates the internal joint contact forces. The inverse dynamics approach may over- or underestimate the resultant external forces and moments compared to the internal joint contact forces estimated by musculoskeletal modeling, which includes the contribution of all muscles spanning a joint [7]. To the best of the authors' knowledge, this is the first systematic study that employed a musculoskeletal model to estimate the knee joint loadings and reports the main muscle forces of the human lower extremity during the full swings of average-skilled golfers with different knee conditions [10]. In addition, the practicability of the model was examined by investigating the biomechanics of the lower extremity of different golfers with and without knee pathologies. In this regard, further studies with a larger cohort of participants should be performed in order to obtain significant clinical data.

Kinematics

The magnitude of the estimated knee flexion angles during the golf swing (57.27–63.20°) is in the same range as reported in previous computational studies [6], [11], [12], [41], but larger than those reported in an *in vivo* study [18] and other computational studies [6], [18], [49]. The range of knee joint mean rotation (approximately 25–31°) is in line with the range of the hip joint rotation [19]. Also, the peak value for mean internal rotation angles of the knee joint (18.3–24.58°) is in agreement with the computational studies (18–26°) of healthy subjects [11], [18], [41], [42], [44], [47], [49]. However, an experimental study with instrumented knee implant reported this value to be 13° at the top of the back swing [46]. The possible reason for this deviation is that the subject of the mentioned study underwent TKA, allowing a maximum of ±12° internal-external rotation, and the knee rotation is structurally restricted by the implant design [46]. However, the TKA subject in the current study showed a peak value of the mean internal rotation close to 25°. Therefore, in the cases of active golf athletes, the type of total knee implant should be thoroughly considered by the surgeon. The high rotation values during golf swing in the knee joint could end in tibiofemoral contacts at the extreme edges of the tibial polyethylene insert surface of fixed-bearing knee endoprosthesis, causing wear at the joint-bearing surface [51]. The mobile-bearing design, on the other hand, provides more freedom of movement, which would reduce polyethylene wear at the tibiofemoral interface in posterior stabilized design, for example [39]. So far, there is no agreement on the implant design that best suits golfers [52]. It may be beneficial to employ knee joint simulators as a valuable experimental tool [27] to help address this question. For this purpose, a combination of position with reaction forces of rotational and translational degrees of freedom of the knee joint, which this musculoskeletal model can provide, may be applied to a six--degrees-of-freedom VIVO™ joint simulator (AMTI, Watertown, MA, USA) to characterize different types of knee endoprosthesis.

Kinetics

Maximum loading or peak value for kinetics of the knee joint and forces of spanning muscle of the knee was observed during the down swing or immediately after the ball impact. The same was reported for hip joint kinetics [17]. Therefore, these can be critical time events during the golf swing, and the golfer should carefully act on these time events, especially those with knee pathologies.

Knee joint moments in the sagittal plane

The peak value of the mean extensor moment of the knee joint for the Healthy subject is (0.73 Nm/kg or 0.042 Nm/BW.Ht, where BW stands for body weight and Ht stands for body height, Fig. 4a, Table 2) comparable to those reported in the simulation study of Choi et al. (0.88 Nm/kg) [12]. However, this is controversially reported in the literature (e.g., 0.021 and 0.025 Nm/BW.Ht [42], 0.08 Nm/BW.Ht [41], and 1.3 Nm/kg [11]). This value appears to be very sensitive to the skill level [12]. This value decreased considerably for the TKA and OA subjects compared to the Healthy subject, which is in line with another study that compares healthy and knee-injured golfers [28]. The same is reported for the level walking and stair walking for the OA subjects [24], as well as stair ambulation [50] and level walking for TKA subjects [36]. This inter-subject deviation may represent compensation by patients to decrease the knee joint loading [24].

Knee joint moments in the transverse plane

The Healthy subject experienced a peak value of mean internal rotation moment (0.21 ± 0.03 Nm/kg) near the BI (Fig. 4(b)) in line with the results from Gatt et al. (0.21 ± 0.13 Nm/kg) [20] and Lin et al. (0.22 ± 0.07 Nm/kg) [28], a little higher than those reported in an experimental study (0.17 ± 0.02 Nm/kg) [14] and another computational study (approximated graphically by 0.15 Nm/kg) [42]. A potential reason for the differences might be the different initial foot rotations. Hooker et al. [21] showed that the initial feet rotation angle influences the peak knee adduction moments, which can also affect the axial rotation moments.

Both patients exhibited considerably decreased peak knee internal rotation moment compared to the Healthy subject. This finding is in accordance with a study on level walking, stair descent and especially stair ascent tasks for TKA subjects [26].

Knee compression force

Musculoskeletal modeling has shown good capability in estimating knee contact forces compared to *in vivo* measurements [31]. However, it has the advantage of non-invasively collecting data from a larger cohort rather than using instrumented implants to measure the loading data in the knee joint.

A peak value of 329% and 303% BW is calculated by the musculoskeletal model for knee compression force for the Healthy and TKA subjects in this study, respectively (Fig. 5), which ties well with a previous experimental study of 320% BW for TKA subject [38] and a computational study calculated a value of approximately 375% BW for a native knee [42].

The knee compression force increased considerably in magnitude by 18.35 (242.4 %), 21.30 (665.6%), 16.14 (138%) N/kg for the TKA, OA, and Healthy subjects from top of back swing to ball impact during the down swing phase, respectively. This is a rapid phase, and a considerable increase in the knee compression force, especially for the OA subject, happened in a concise time period that can be a potential negative mechanical factor causing long-term knee prognosis such as worsening an OA knee or loosening knee endoprosthesis. It was shown that the knee joint loading rate is strongly connected to joint degeneration [37], and the rate of loading of a joint has been hypothesized as a significant component in the advancement of OA [15].

Muscle forces

The musculoskeletal model can provide all lower extremity muscle forces during the golf swing, which is hardly viable in experiments.

Knee extensor muscles (m. rectus femoris, m. vastus lateralis) showed their peak muscle forces during the forward swing phase, most likely to position the knee above the lead foot to prepare and support it to receive the high knee compression forces near the ball impact in acceleration or early follow through phases [34]. Comparison between peak values of knee extensor muscle forces (Table 4) shows that m. vastus lateralis provides 71.97% more force than the m. rectus femoris for the Healthy subject. This can lead to the assumption that m. vastus lateralis contributes more in providing the lead knee extension moment during the golf swing. In addition, the reduction in knee extension moment for patients compared to the Healthy subject (Fig. 4a) is in line with this assumption because the peak value for m. vastus lateralis force is considerably reduced for patients compared to the Healthy subject (Figs. 6c, 6l). Therefore, it can be assumed that m. vastus lateralis might play a more critical role in lead knee extension moment generation than m. rectus femoris during the golf swing. Since the peak knee extension moment happens prior to the ball impact, it can be concluded that strengthening the m. vastus lateralis of the lead knee as a knee extensor might be of great benefit to have better golf shots. The reduction in peak m. vastus lateralis force for patients is in accordance with experimental studies, which revealed that the knee extensor muscle strength is significantly decreased in TKA and OA patients [40]. A musculoskeletal simulation reported the same result for stair ambulation of TKA subjects [45].

M. biceps femoris and m. gastrocnemius produced the largest knee flexor muscle forces (Figs. 6f, g). However, these muscles produced the peak muscle forces at different time points. This was observed after the ball impact at the beginning of the early follow through for m. biceps femoris and before the ball impact at the beginning of the acceleration phase for m. gastrocnemius. Since it happens before ball impact for m. gastrocnemius, as a knee flexor, it might contribute more to generating a higher performance swing. Opposed to the stair ambulation [45], the peak value for m. gastrocnemius force during golf was considerably decreased for the patients (Fig. 6g). In this regard, Astephen et al. [5] reported a decreased medial m. gastrocnemius activity for severe OA subjects compared to the moderate one while walking, which might be an effort to reduce knee joint loading and pain by OA subjects [22].

Limitations

Besides the advantages of our approach, the present study has some limitations. The main concern about findings from the inter-subject comparisons in this study is the minimum homogenous sample size. The main intention of this comparison study was to develop the musculoskeletal model for different types of subjects and assess its efficacy in capturing inter-subject differences. Thus, the reader should interpret the findings of the practicability comparison study with care. Another limitation involves the issue of using a unique musculoskeletal model for Healthy subjects and TKA or OA patients. This assumes the characteristics of the muscles, e.g., geometry and strength, is the same following a total knee arthroplasty and for an OA knee [4]. An apparent limitation of the lower extremity model in this study is the idealized topology of the knee joint model considering only two degrees of freedom. In this regard, more complex knee joint models such as force-dependent kinematics [3] or other advanced models presented in [25] may offer additional benefits in future studies.

5. Conclusions

In this study, we presented and evaluated an adapted musculoskeletal model that can directly integrate subject-specific body height, body weight, motion-captured data, and ground reaction forces to estimate the knee joint loadings and muscle forces of the lower extremity during the golf swing. The simulation results were comparable to previous experimental and computational studies. Therefore, the model can be regarded as a valuable tool to assist clinicians and scientist in obtaining additional insights into musculoskeletal biomechanics of golf, e.g., evaluating the joint range of motion and performance, e.g., after total knee arthroplasty or to better understand the injury occurrence by estimating the lower extremity loadings during golf swings. Future research should focus on improving the complexity and personalization of the musculoskeletal model, as well as integration of a larger cohort of subjects to strengthen the outcomes of our practicability study.

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Conflict of interest

The authors report that there are no competing interests to declare.

Ethical approval

This study was approved by the local ethics committee of the Rostock University Medical Center (ID: A2020-0191).

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