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A chain kinematic model to assess the movement of lower-limb including wobbling masses.

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Abstract

Soft tissue and bones kinematics can only accurately be distinguished using invasive methods. We propose a non-invasive method to estimate kinematics of both parts in the lower-limb. Modelled as a chain model, including rigid and wobbling bodies, its states were determined using a least-square algorithm. By comparison to a classical model without wobbling bodies, the marker residual was reduced by 20% but the joint kinematics remains unchanged. Average wobbling bodies displacements reached $6.9 \pm 3.5^\circ$ and 6.9 ± 2.4 mm relative to the modeled bones. This original method is a first step to assess the effect of wobbling masses on joint kinetics.

Keywords: Wobbling masses; Kinematics; Global Optimization; Joint Torques

Introduction

In human movement analysis, joint torques are calculated using inverse dynamics in which each body segment is assumed to be rigid. Since body segments can deform (Pain and Challis 2006), the rigidity assumption introduces some bias into joint torque estimates. Indeed, two-dimensional computer simulation models with wobbling masses showed that joint efforts (*e.g.* joint torques and joint forces) were 50% lower than those estimated with rigid bodies only (Gruber, Ruder et al. 1998, Gittoes, Brewin et al. 2006, Pain and Challis 2006). At impact, the deceleration of the wobbling masses was lower and occurred later than the bone deceleration (Challis and Pain 2008). To our knowledge, computer simulations remain the only method to have assessed the wobbling masses effect on joint dynamics.

Previous models defined each body segment with two rigid parts, namely bone and wobbling masses that move with respect to the bones with three degrees of freedom (*DoF*), namely translations along the antero-posterior and longitudinal axes and rotation around the medio-lateral axis of the underlying bones (Gruber, Ruder et al. 1998, Gittoes, Brewin et al. 2006). However an in-vivo study (Gao and Zheng 2008) has shown that skin marker triads mainly rotate around the longitudinal axis of the thigh and shank. Results of simulation models based on planar models could therefore be biased and thus requiring experimental three-dimensional data.

Experimental studies do not separate the bone and wobbling mass kinematics except in intra-cortical pin based experiments. Such experiments have shown that skin markers had a movement composed of a random, marker-specific trajectory (termed as marker deformation), and a common movement called rigid transformation of markers (RTM) caused by the subjacent wobbling masses (Leardini, Chiari et al. 2005). In-vivo and ex-vivo studies using intra-cortical pins have shown that RTM is greater than marker deformation (Andersen, Damsgaard et al. 2012, Grimpampi, Camomilla et al. 2012). Such results suggest that the assessment of the RTM with respect to the bone can be used to determine wobbling mass kinematics.

Experimentally, only intra-cortical pins, in addition to skin markers, could estimate the soft tissue kinematics with respect to the bones. However, they are invasive and provoke pain, especially in tasks involving large impacts with the ground and their use is limited to a few laboratories since their insertion requires sterile conditions (Ramsey, Wretenberg et al. 2003). Pins also constrain the wobbling mass motion and sometimes bend due to tendon and muscle contact forces. All these drawbacks have led to the development of mathematical methods to better estimate bone kinematics by minimizing the effect of soft tissue artefacts (STA) (Leardini, Chiari et al. 2005).

Two kinds of mathematical methods have been developed to minimize the effect of STA. *Segmental optimization* methods consider each body segment independently with six *DoF* and minimize the deformation of its cluster of markers (Alexander and Andriacchi 2001, Taylor, Ehrig et al. 2005). Such an assumption can lead to non-physiological joint translation (Leardini, Chiari et al. 2005). To overcome this limitation, *multibody optimization* methods include all segments in a chain model with set *DoF*. The chain model configuration is obtained by minimizing all distances between actual and modelled markers (Lu and O'Connor 1999). The segmental lengths remain constant and “represent the bones” (Ojeda, Martínez-Reina et al. 2014). Some studies have reported better joint kinematics using a multibody-optimization rather than a segmental optimization approach (Klous and Klous 2010, Joao, Veloso et al. 2014). Other studies have introduced a soft tissue artefact model in the multibody optimization, which takes into account RTM and does not affect joint kinematics (Richard, Camomilla et al. 2012). However, they did not assess the wobbling mass kinematics and their contribution to joint torques remains unknown.

Assuming that a chain model represents the bones, the inclusion of a body representing the wobbling masses in the chain model as in Pain and Challis (2006) should provide a non-invasive method to assess the wobbling mass kinematics. Such a model would be a first step to experimentally investigate the

effect of wobbling masses on joint kinetics. The aim of this study was (1) to propose a chain model of the lower-limb including wobbling masses in order to assess their movement. The RTM was considered as the wobbling mass movement and (2) used to determine their main axis of movement. The inclusion of wobbling bodies in a chain model could alter joint kinematics. Hence, the new chain model must estimate the joint kinematics similarly to classical chain models.

Material and Methods

Experimental protocol

Eighteen subjects (23.3 ± 3.4 years old, 1.81 ± 0.11 m, 75.2 ± 10.4 kg, 22.9 ± 2.8 m.kg⁻² BMI) took part in this study after giving their informed consent. Eighty-five markers were placed on the lower-limb: pelvis (7), thigh (32), shank (27) and foot (19) (Figure 1a). For the pelvis, the markers were put on the two anterior inferior and superior iliac spines, on the two posterior iliac spines and on the posterior spine of the fifth sacral vertebra. For the thigh, the markers were distributed in six lines of equidistant markers using a tape measure (Roosen, Pain et al. 2013) to ensure a reproducible marker placement between subjects. This procedure was repeated for both the shank and the foot.

Please insert Figure 1 here

First, the subjects were instructed to stand still 5 s in an anatomic posture to get the reference position of the markers with respect to their body segment. Subjects then performed functional movements in order to locate the hip and ankle joint centres and the knee flexion axis. Finally, a hopping trial composed of 20 jumps at a self-selected frequency and height of jump was recorded. Subjects received the instruction to bounce with the knees as straight as possible. A force plate was used to detect the landing and take-off times of the jump series. The hopping movement was chosen because it limits flexion at the knee joint and consequently the coupling motions, i.e. translations, adduction and internal rotation

(Belvedere, Leardini et al. 2011). All marker trajectories were acquired using a 14-camera-optoelectronic system (Vicon, Oxford, 40 Mpx) sampled at 300 Hz according to the recommendations of Pain and Challis (2006).

Based on all the markers, hip and ankle joint centres were located using the SCoRE algorithm (Ehrig, Taylor et al. 2006). The knee flexion joint axis was determined using the SARA functional method (O'Brien, Bodenheimer et al. 2000). The knee joint centre was then defined as the projection on the flexion axis of the mid-point of markers placed on the condyles. The axes of the segmental system of coordinates (SOC) were built to correspond with the anatomical axes (X: medio-lateral, Y: antero-posterior, Z: longitudinal) and their origin was the proximal joint center. The location of each marker is expressed in the SOC of the corresponding body segment and used to determine the state of the chain model.

Lower limb models

Two chain models were built to represent the lower-limb (Figure 1b, Table 1). The first one, called Rigid Model (RM) was a 13 *DoF* chain model composed of four rigid bodies representing the pelvis, thigh, shank and foot. These bodies were linked by the hip, the knee and the ankle respectively modelled as ball, hinge and ball joints respectively. For the knee, the abduction-adduction and internal-external rotations were considered fixed during the movement. These angles were determined during the static trial using the orientation of the thigh and shank segmental frames for both models.

The second chain model had 25 *DoF*, named as Wobbling Model (WM). This model added two wobbling bodies to the RM, representing the soft tissues overlying the thigh and the shank with 6 *DoF* joints for both bodies. The chain models details are summarized in Table 1.

Please insert Table 1 here

The two chain models were reconstructed through a multibody optimization algorithm using a quadratic algorithm similar to Begon, Wieber et al. (2008) and (Fohanno, Lacouture et al. 2013).. The optimal pose (\mathbf{q}) of the chain model was obtained for each data frame by minimizing the least squared distance between the actual and the modeled markers in a Newton-Gauss algorithm. The main difference in the model reconstruction between the RM and WM is related to the number of markers involved in the reconstruction process. Indeed, the rigid part of the WM (femur and tibia) is determined using only 36 markers corresponding to those placed on the pelvis, the medial face of the tibia and those placed on the foot. For the WM model, this marker selection means that the marker placed on the thigh and those placed on the posterior face of the shank are used to determine the wobbling mass kinematics. This can be done as the Hessian matrix involved in the Newton Gauss algorithm still remains full ranked.

Data analysis

For the hopping trial, each of twenty jumps was defined as the frame of events between two successive landings. The latter comprised the landing, contact, take-off and mid-aerial phases. Each jump was time-normalized to 200 frames (i.e. 100 frames for contact and aerial phase each). The mean and standard deviation of the joint and wobbling mass kinematics were computed on the determined average-jump.

Model comparisons

The two models were compared using the marker residual to assess the reconstruction quality. This parameter is defined as the mean difference between the measured and the model-determined markers across the entire trial. The difference between the two models was assessed using the Friedman test.

The joint kinematics of RM and WM were compared with regards to joint angles and joint translations.

The RM was used to verify if the joint kinematic obtained was not jeopardized when using the WM

model. For each subject, the *RMSd* was calculated for each joint angle (hip, knee, ankle) and segment translation in the global frame called T_x , T_y and T_z .

A two-way ANOVA with repeated measures was used to compare the models. The second factor corresponded to the four different events of the jump, namely landing, the middle of the contact phase, the take-off and the middle of the aerial phase. If the joint kinematics were similar, then the RTM was analyzed to determine the main axis of the wobbling mass kinematics.

Rigid transformation kinematics

The rotations observed around the medio-lateral, antero-posterior and longitudinal axes were named backward and forward swing, medial and lateral tilt and internal and external twist, respectively. The translations were named as shifts in the antero-posterior, medio-lateral and proximo-distal direction. The amplitude of rotation and translation on each axis was reported in order to determine the axis with the most motion in rotations and translations. These values were compared using a Kruskal-Wallis test to detect if there were a main axis of translation and rotation. If significant, the Mann-Whitney-Wilcoxon post-hoc test was applied to determine the main axis. All statistics tests were made using IBM SPSS Statistics 19 with $\alpha = 0.05$, for the kinematic comparisons between the two models and for the determination of the main axis.

RESULTS

The mean RM marker residual is equal to 6.5 ± 2.2 mm whereas the mean WM marker residual is equal to 5.2 ± 0.9 mm. The WM model significantly reduced the marker residual of 20% in regards with marker residual obtained using the RM model ($p = 0.01$).

Joint Kinematics

The *RMSd* between the two models ranged from $0.4 \pm 0.4^\circ$ to $2.1 \pm 1.4^\circ$ for the joint angles and from 0.7 ± 0.6 mm to 3.5 ± 1.4 mm for the segment translations in the global frame (Table 2). No significant

differences were revealed between the two chain models for all joints, rotations and positions involved ($p>0.05$). For all joints, the mean and standard deviations of joint kinematics for all subjects are illustrated in Appendix A1.

Please insert Table 2 here

Rigid transformation kinematics

The average amplitudes of the thigh wobbling masses rotations and translations among the 18 subjects ranged from $1.2\pm 0.5^\circ$ to $7.9\pm 3.9^\circ$ and from 6.2 ± 4.0 to 11.2 ± 4.4 mm, respectively. The statistical analysis demonstrated that the wobbling masses mainly turn around the longitudinal axis and slide along the antero-posterior axis of the thigh (Table 3). Figure 2 shows these results for one subject.

Please insert Table 3 here

Please insert Figure 2 here

Regarding the shank, the mean amplitude in rotation of the wobbling masses ranged from $0.9\pm 0.5^\circ$ to $6.9\pm 3.5^\circ$ and the translations ranged from 5.4 ± 2.1 mm to 6.9 ± 2.4 mm. The statistical analysis demonstrated that the wobbling masses of the shank mainly turn around the longitudinal axis and slide along the antero-posterior axis (Table 3). These results for another subject are represented in Figure 3.

Please insert Figure 3 here

The mean and standard deviation of the wobbling mass kinematics observed during the average jump among the 18 subjects are available in Appendix A2 and A3 respectively for the thigh and the shank. Due to high standard deviations, no pattern of the wobbling mass movement among subjects can be

defined, suggesting that the movement is subject-specific. Although, for both segments the statistical analysis shows that the wobbling masses mainly turn around the longitudinal axis and mainly slide along the antero-posterior axis.

DISCUSSION

This study is a first attempt to dissociate lower-limb bone and wobbling mass kinematics using skin markers only. This original method consisted in modeling the rigid and wobbling (6 *DoF*) components of the lower-limb in a multibody chain and estimating the orientation of the femur without taking into account the skin markers located on the thigh. The latter were used to determine the position and orientation of the thigh with respect to the femur only. The main findings are that the WM model gave similar joint kinematics to a classical multibody kinematic approach (Lu and O'Connor 1999) and that wobbling masses mainly turn around the longitudinal axis and slide along the antero-posterior axis of both the femur and tibia during hopping. Similarly to the non-invasive methods developed to assess STA (de Rosario, Page et al. 2012, Monnet, Thouzé et al. 2012), the model will have to be validated to estimate its accuracy. Intra-cortical pin based kinematics could be the gold standard measure to validate the model, as only one joint can be study using the bi-planar fluoroscopy (Leardini, Chiari et al. 2005, Barre, Thiran et al. 2013). However, a restricted sample must be expected (Peters, Galna et al. 2010).

Model effects

Several studies comparing models demonstrated that the most biofidelic models gave the lowest reconstruction errors (i.e. marker residual) without requiring additional *DoF* (Fohanno, Lacouture et al. 2013, Laitenberger, Raison et al. 2014). In the present study, the WM model also ensured a higher quality of reconstruction by dividing the body segments in two different bodies giving more *DoF* to the WM model. Its bio-fidelity could be improved using a knee parallel mechanism taking into account the

knee joint rotations and translations in function of the flexion-extension joint angle (Duprey, Cheze et al. 2010). However the loop closure constraints have to be adjusted so that the final number of *DoF* at the knee remains equal to 1. The estimation of the femur orientation with respect to the pelvis depends on this assumption, otherwise the Hessian matrix in the Newton-Gauss algorithm is not anymore of full rank.

Model parameters, such as joint type or marker sets associated to each body segment, are known to significantly affect joint kinematics (Duprey, Cheze et al. 2010). While no marker was attached directly to the femur, the inclusion of two wobbling bodies representing the soft tissues package had no significant effect on joint kinematics during hopping in a population of 18 young healthy and normal weighted participants. This result fits with the previous work of Richard, Camomilla et al. (2012) in which a STA model in a multibody optimization was included without seeking to estimate wobbling mass state. Similar to Schulz and Kimmel (2010), the WM model also revealed that the tibia and pelvis kinematics are sufficient driving constraints to estimate the femur orientation in dynamics tasks as no significant difference were highlighted in joint kinematics for both studies. In summary, the model proposed in this study can estimate joint kinematics whilst minimizing the reconstruction error in comparison with a classical chain model and thus could be recommended to estimate the wobbling mass movement.

Wobbling mass movement

In hopping with relatively straight legs, wobbling masses mainly rotated around the longitudinal axis and slid along the antero-posterior axis of both the femur and tibia. These results are in agreement with those of Gao and Zheng (2008) who also used skin markers but quantified STA by measuring inter-marker translations and rotations during walking trials. This suggests that future computer simulation models built for assessing the wobbling mass effects on joint kinetics should consider these axes of

movement which may exert a shear force on the joints. Such aspect will require further evaluation through future research.

While the overall joint kinematics was similar between the two models, a 1-DoF knee joint could bias the wobbling mass kinematics (Andersen, Benoit et al. 2010). The values of the rotation and translation of the shank wobbling masses must be considered by the model's assumptions. The angles of abduction-adduction and internal and external rotation were modeled as constants and no joint translations allowed whereas these coupling motions vary with the flexion angle (Belvedere, Leardini et al. 2011). However the range of flexion-extension ($>50^\circ$) during hopping stayed far from the values where the knee joint translation is significant according to Belvedere, Leardini et al. (2011) and Myers, Torry et al. (2011). This reinforces the validity of the model but could limit its application to other movements with greater knee flexion such as running. The tilt and twist movement of the wobbling masses might however be overestimated by this noninvasive approach.

The high between-subjects variability of the wobbling mass movement highlighted that these movements are subject-specific. Furthermore, the difference of mass distribution between subjects (Schinkel-Ivy, Burkhart et al. 2012) and the impact with the ground (Pain and Challis 2002) have been considered as the principal sources of variability. A limitation to our study was to not standardize the jump height, which is another source of variability. Additionally, the strategy of muscles recruitment could have affected the wobbling mass kinematics. Accordingly, Boyer and Nigg (2006) observed that subjects with more toned muscles exhibited less soft tissue vibrations. Future studies should correlate wobbling mass kinematics to other biomechanical factors.

The wobbling masses can exhibit different decelerations as a function of their placement on the segment (Pain and Challis 2006). The present study cannot inform which part of the body segment moved the most because the WM chain model associated a single wobbling mass to each bone. The priority for future work would be to introduce the inertial parameters of the different bodies, which remains challenging (Challis and Pain 2008). Such an estimation could partially rely on the regression equations

of Holmes, Andrews et al. (2005) to differentiate bone and soft tissue masses. Then the model could be applied to assess the contribution of the wobbling masses in the knee shear forces, which are related to ligament injuries, using an inverse dynamic approach.

In conclusion, our original method consists in modeling the rigid and wobbling parts of the segments in a chain model reconstructed using a multibody optimization algorithm. This chain model is able to assess the movement of the wobbling masses in regard to modeled bones without jeopardizing joint kinematics under the assumption of a 1-DoF knee model. The wobbling masses of the thigh and shank mainly turn around the longitudinal axis and slide along the antero-posterior axis of their respective modeled bone. This non-invasive method is a prerequisite to future studies directed at experimentally assessing the effect of the wobbling masses in joint kinetics.

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References

- Alexander, E. J. and T. P. Andriacchi (2001). "Correcting for deformation in skin-based marker systems." Journal of Biomechanics **34**(3): 355-361.
- Andersen, M. S., D. L. Benoit, M. Damsgaard, D. K. Ramsey and J. Rasmussen (2010). "Do kinematic models reduce the effects of soft tissue artefacts in skin marker-based motion analysis? An in vivo study of knee kinematics." Journal of Biomechanics **43**(2): 268-273.
- Andersen, M. S., M. Damsgaard, J. Rasmussen, D. K. Ramsey and D. L. Benoit (2012). "A linear soft tissue artefact model for human movement analysis: proof of concept using in vivo data." Gait and Posture **35**(4): 606-611.
- Barre, A., J. P. Thiran, B. M. Jolles, N. Theumann and K. Aminian (2013). "Soft tissue artifact assessment during treadmill walking in subjects with total knee arthroplasty." IEEE Trans Biomed Eng **60**(11): 3131-3140.
- Begon, M., P. B. Wieber and M. R. Yeadon (2008). "Kinematics estimation of straddled movements on high bar from a limited number of skin markers using a chain model." Journal of Biomechanics **41**(3): 581-586.
- Belvedere, C., A. Leardini, S. Giannini, A. Ensini, L. Bianchi and F. Catani (2011). "Does medio-lateral motion occur in the normal knee? An in-vitro study in passive motion." Journal of Biomechanics **44**(5): 877-884.
- Boyer, K. A. and B. M. Nigg (2006). "Soft tissue vibrations within one soft tissue compartment." Journal of biomechanics **39**(4): 645-651.
- Challis, J. H. and M. T. G. Pain (2008). "Soft tissue motion influences skeletal loads during impacts." Exercise and Sport Sciences Reviews **36**(2): 71-75.
- de Rosario, H., A. Page, A. Besa, V. Mata and E. Conejero (2012). "Kinematic description of soft tissue artifacts: quantifying rigid versus deformation components and their relation with bone motion." Medical & Biological Engineering & Computing **50**(11): 1173-1181.
- Duprey, S., L. Cheze and R. Dumas (2010). "Influence of joint constraints on lower limb kinematics estimation from skin markers using global optimization." Journal of Biomechanics **43**(14): 2858-2862.
- Ehrig, R. M., W. R. Taylor, G. N. Duda and M. O. Heller (2006). "A survey of formal methods for determining the centre of rotation of ball joints." Journal of Biomechanics **39**(15): 2798-2809.
- Fohanno, V., P. Lacouture and F. Colloud (2013). "Improvement of upper extremity kinematics estimation using a subject-specific forearm model implemented in a kinematic chain." Journal of Biomechanics **46**(6): 1053-1059.
- Gao, B. and N. Zheng (2008). "Investigation of soft tissue movement during level walking: Translations and rotations of skin markers." Journal of Biomechanics **41**(15): 3189-3195.
- Gittoes, M. J., M. A. Brewin and D. G. Kerwin (2006). "Soft tissue contributions to impact forces simulated using a four-segment wobbling mass model of forefoot-heel landings." Human Movement Science **25**(6): 775-787.
- Grimpampi, E., V. Camomilla, A. Cereatti and A. Cappozzo (2012). "Quantative assessment of soft tissue artefact propagation to the marker cluster level." Journal of Biomechanics **45**: S296.
- Gruber, K., H. Ruder, J. Denoth and K. Schneider (1998). "A comparative study of impact dynamics: wobbling mass model versus rigid body models." Journal of Biomechanics **31**(5): 439-444.
- Holmes, J. D., D. M. Andrews, J. L. Durkin and J. J. Dowling (2005). "Predicting in vivo soft tissue masses of the lower extremity using segment anthropometric measures and DXA." Journal of applied biomechanics **21**(4): 371-382.

Joao, F., A. Veloso, S. Amadao, P. Armada-Da-Silva and A. C. Mauricio (2014). "Can global optimization technique compensate for marker skin movement in rat kinematic?" Journal of Mechanics in Medicine and Biology In Press.

Klous, M. and S. Klous (2010). "Marker-based reconstruction of the kinematics of a chain of segments: a new method that incorporates joint kinematic constraints." Journal of Biomechanical Engineering **132**(7): 074501.

Laitenberger, M., M. Raison, D. Périé and M. Begon (2014). "Refinement of the upper limb joint kinematics and dynamics using a subject-specific closed-loop forearm model." Multibody System Dynamics: 1-26.

Leardini, A., L. Chiari, U. Della Croce and A. Cappozzo (2005). "Human movement analysis using stereophotogrammetry - Part 3. Soft tissue artifact assessment and compensation." Gait and Posture **21**(2): 212-225.

Lu, T. W. and J. J. O'Connor (1999). "Bone position estimation from skin marker co-ordinates using global optimisation with joint constraints." Journal of Biomechanics **32**(2): 129-134.

Monnet, T., A. Thouzé, M. T. G. Pain and M. Begon (2012). "Assessment of reproducibility of thigh marker ranking during walking and landing tasks." Medical Engineering & Physics **34**(8): 1200-1208.

Myers, C. A., M. R. Torry, D. S. Peterson, K. B. Shelburne, J. E. Giphart, J. P. Krong, S. L. Woo and J. R. Steadman (2011). "Measurements of tibiofemoral kinematics during soft and stiff drop landings using biplane fluoroscopy." The American Journal of Sports Medicine **39**(8): 1714-1722.

O'Brien, J. F., R. E. Bodenheimer, G. J. Brostow and J. K. Hodgins (2000). Automatic Joint Parameter Estimation from Magnetic Motion Capture Data. Graphics Interface, Montréal, Qc, Canada.

Ojeda, J., J. Martínez-Reina and J. Mayo (2014). "A method to evaluate human skeletal models using marker residuals and global optimization." Mechanism and Machine Theory **73**(0): 259-272.

Pain, M. T. and J. H. Challis (2002). "Soft Tissue Motion During Impacts: Their Potential Contributions to Energy Dissipation." Journal of Applied Biomechanics **18**(3).

Pain, M. T. and J. H. Challis (2006). "The influence of soft tissue movement on ground reaction forces, joint torques and joint reaction forces in drop landings." Journal of Biomechanics **39**(1): 119-124.

Peters, A., B. Galna, M. Sangeux, M. Morris and R. Baker (2010). "Quantification of soft tissue artifact in lower limb human motion analysis: a systematic review." Gait and Posture **31**(1): 1-8.

Ramsey, D. K., P. F. Wretenberg, D. L. Benoit, M. Lamontagne and G. Nemeth (2003). "Methodological concerns using intra-cortical pins to measure tibiofemoral kinematics." Knee Surgery Sports Traumatology Arthroscopy **11**(5): 344-349.

Richard, V., V. Camomilla, L. Chèze and R. Dumas (2012). Introducing a soft tissue artifact model in multi-body optimization: a feasibility study. 3-D Analysis of Human Movement, San Fransisco, United States.

Roosen, A., M. T. Pain, A. Thouze, T. Monnet and M. Begon (2013). "Segment-embedded frame definition affects the hip joint centre precision during walking." Med Eng Phys.

Schinkel-Ivy, A., T. A. Burkhart and D. M. Andrews (2012). "Leg tissue mass composition affects tibial acceleration response following impact." Journal of Applied Biomechanics **28**(1): 29-40.

Taylor, W. R., R. M. Ehrig, G. N. Duda, H. Schell, P. Seebeck and M. O. Heller (2005). "On the influence of soft tissue coverage in the determination of bone kinematics using skin markers." Journal of Orthopaedic Research **23**(4): 726-734.

Tables

Table 1. Characteristics of the RM and WM chain models with body segments and their corresponding markers (see Fig. 1)

Chain Model	Segment Number	Name	Parent	Degree of freedom	Markers
Rigid Model	S1	Pelvis	Global	q_{1-6}	1-7
	S2	Thigh	S1	q_{7-9}	8-39
	S3	Shank	S2	q_{10}	40-66
	S4	Foot	S3	q_{10-13}	67-85
Wobbling Model	S1	Pelvis	Global	q_{1-6}	1-7
	S2	Thigh-bone (TB)	S1	q_{7-9}	None
	S3	Shank-bone (SB)	S2	q_{10}	40-52
	S4	Foot	S3	q_{10-13}	67-85
	S5	Thigh-Mass (TM)	S2	q_{14-19}	8-39
	S6	Shank-Mass (SM)	S3	q_{20-25}	53-66

Table 2. Mean and standard deviation of RMS differences between the two chain models in terms of joint kinematic and segment positions in the global frames

Joint	Rotations (°)			Translations (mm)		
	Flexion- Extension	Abduction- Adduction	Internal- External	Medio- Lateral	Antero- Posterior	Vertical
Hip	0.9±0.7	1.0±0.5	2.1±1.4	3.0±1.1	2.4±2.0	0.8±0.7
Knee	0.9±0.4			1.8±1.9	3.5±1.4	0.7±0.6
Ankle	0.4±0.4	1.5±1.2	1.2±0.9	2.3±1.4	1.2±0.8	1.6±0.6

Table 3 : Mean and standard deviation, minimal and maximal amplitudes of the wobbling mass movements for the thigh and the shank wobbling masses. The asterisk highlights the main axis of movement

		Rotation			Translation		
		Swing (°)	Tilt (°)	Twist (°)	Medio- Lateral (mm)	Frontward- Backward (mm)	Upward- Downward (mm)
Thigh	Mean	1.9±0.8	1.2±0.5	7.9±3.9*	6.2±4.0	11.2±4.4*	6.3±3.6
	Min	1.1	0.5	4.5	2.7	7.4	3.6
	Max	3.4	1.8	21.1	11.1	22.7	12.9
Shank	Mean	1.6±0.4	0.9±0.5	6.9±3.5*	5.4±2.1	6.9±2.4*	6.1±2.4
	Min	1.4	0.4	2.7	2.1	5.2	3.4
	Max	2.5	1.8	12.4	12.9	11.4	7.6

Figure 1. a) Front and back views of the marker placement on the left lower-limb; b) Chain models: Rigid Model (RM) in full line and the two additional wobbling bodies in dotted line which compose the Wobbling Model (WM).

Figure 2. Mean kinematic (black line) and standard deviation (grey area) of S5 thigh wobbling masses during the average jump. LA and TO represent landing and takeoff events respectively.

Figure 3. Mean kinematic (black line) and standard deviation (grey area) of the S10 shank wobbling masses during the average jump. LA and TO represent landing and takeoff events respectively.

Appendix

The mean joint kinematics and standard deviation among the subjects obtained with the two models are available in the Figure A1. Figure A1 shows that the two joint kinematics overlap during the average jump. The constant angles for the knee abd-adduction and internal-external rotation are the consequence of the only degree of freedom.

Please insert Figure A1 here

Figure A1. Mean and standard deviation of joint angles for the RM (line) and WM (dotted line) chain models from landing (LA) to take-off (TO) and aerial phase from TO to LA.

Figure A2 and A3 show respectively the thigh and shank wobbling mass kinematics in regard with their respective modeled “bones”. In order to illustrate the main axis of the wobbling mass movements in rotation and translation, the same scale was applied for the rotation and translation.

Please insert Figures A2 and A3 here

Figure A2 : Mean and standard deviation of the thigh wobbling masses in the anatomical frame of the thigh during the average jump of all subjects.

Figure A3. Mean and standard deviation time histories of the shank wobbling mass kinematics in the anatomical frame of the shank during the average jump of all subjects.