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ISB recommendations on the reporting of intersegmental forces and moments during human motion analysis

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Locomotion, gait, methods, torque, biomechanics

Abstract

The International Society of Biomechanics (ISB) has charged this committee with development of a standard similar in scope to the kinematic standard proposed in Wu et al., 2002 and Wu et al., 2005. Given the variety of purposes for which intersegmental forces and moments are used in biomechanical research, it is not possible to recommend a particular set of analysis standards that will be acceptable in all applications. Instead, it is the purpose of this paper to recommend a set of reporting standards that will result in an understanding of the differences between investigations and the ability to reproduce the research. The end products of this standard are 1) a critical checklist that can be used during submission of manuscripts and abstracts to insure adequate description of methods, and 2) a web based visualization tool that can be used to alter the coordinate system, normalization technique and internal/external perspective of intersegmental forces and moments during walking and running so that the shape and magnitude of the curves can be compared to one's own data.

Introduction

Progress in any field of inquiry relies on the ability of researchers to compare previously published results and replicate research. As complexity of design and analysis increases this becomes more challenging. The nature of human motion research is such that direct measurement techniques are rarely available and often inadequate to measure internal loading during activities of daily living and exercise. We often rely on layers of models to estimate these loads and apply the models in a variety of ways. Results from in-silico, i.e. computer based simulations, in-vitro, i.e. anatomical specimens, and in-vivo measures are produced in specific research centers, but then reported at national and international levels, in congresses and in

journals, to be shared within the scientific community. There is a need to establish a shared knowledge base, to benefit populations of interest, and ultimately to improve the life of individuals (patients, athletes, workers, etc.). In order to effectively communicate the results of these studies, calculations must be done correctly and reported clearly, with the goal that the research can be understood and replicated without ambiguity. Relevant dissemination of results must be according to standard mechanics, consistent with human body anatomy, and comprehensible by any professional involved, no matter the medical, engineering, technical or industrial background of the reader. In our field of study confusion exists on these matters, with evident errors in a number of published papers, and incomprehension and questionable interpretation of many available results. This hinders the ability of researchers to take advantage of the shared knowledge base. A number of review papers have investigated explicit protocols and techniques for human motion analysis, but only a few specific research topics such as finite element modelling (Burkhart et al. 2013), multi-segment foot kinematics analysis (Bishop et al. 2012) and soft tissue artefact description (Cereatti et al., 2017) have received recommendations.. In this regard, the International Society of Biomechanics (ISB) attempted standardization for the description of joint kinematics in two papers (Wu et al. 2002, 2005), which have received more than 1300 and 1600 citations respectively (as of March 2019, Scopus).

Fundamental quantities of interest in human motion research are the intersegmental forces and moments acting at the joints. These forces and moments represent the net loads that act at a joint. The resultant forces should not be considered physical interactions that occur within the joint as they are often many times smaller than the actual joint contact forces, which include the contribution of muscles (Scott and Winter, 1990). Both force and moment vectors are usually decomposed into three components and transformed into a relevant three-dimensional

coordinate system for presentation purposes. This can be accomplished by projecting the vectors onto the corresponding axes. These axes will be referred to as the superior-inferior, anterior-posterior and medial-lateral axes. Intersegmental moments can be referred to by their action: internal-external rotation, adduction-abduction and flexion-extension; the plane in which the moment acts: transverse, frontal and sagittal; or by the axis of rotation: superior-inferior, anterior-posterior and medial-lateral. Intersegment moments can be analysed on their own or used in the further estimation of muscle forces and joint contact forces. However, intersegmental forces are not the total force acting at a joint and therefore have limited utility on their own (except for specific cases such as kinetic analysis in the prosthetic joints in amputees, Dumas et al., 2017) but are necessary for the estimation of joint moments and joint contact forces.

There are a number of decisions that need to be made in the collection and analysis stages and these must be described in any dissemination stage because they affect the calculated values and the interpretation of the results. Among these are the anthropometric modelling, joint center estimation, smoothing/filtering, method of calculation, coordinate system, evaluation perspective (internal or external), and normalization. As an example of the inherent variety of results different methodological choices can make, sagittal plane knee joint moments during walking are presented in Figure 1 from eight research studies on healthy adults. It is presumed that these curves were all calculated correctly yet various methods and coordinate systems were utilized, thus altering the shape of the curves. Failure to adequately describe the methods will result in data that cannot be interpreted by the reader nor replicated by the research community.

The aims of the present paper are to discuss the major issues in the definition, calculation, and interpretation of intersegmental forces and moments in human motion analysis, and to make final recommendations on these matters with guidance from relevant papers in the literature. The

goal is to eliminate the most frequent sources of error and confusion in the field of human motion analysis so that research can be correctly interpreted and replicated. We are not putting forth these recommendations in an attempt to standardize the methods of estimating intersegmental forces and moments, rather we hope that it is seen as an attempt to standardize the reporting of such methods, after careful consideration of procedures and calculations have been applied.

Anthropometric Model

The relationships between kinetic variables (force and moment) and kinematic variables (linear and angular velocity and acceleration) are governed by the anthropometric properties of mass and moment of inertia about the center of mass.

$$\sum \mathbf{F} = \mathbf{m}\mathbf{a}$$

where, $\sum \mathbf{F}$ is the sum of the external forces applied to a given human body segment

\mathbf{m} is the segment mass,

\mathbf{a} is the linear acceleration of the center of mass.

$$\sum \mathbf{M} = \mathbf{I}^{cm} \dot{\boldsymbol{\omega}} + \boldsymbol{\omega} \times \mathbf{I}^{cm} \boldsymbol{\omega}$$

where, $\sum \mathbf{M}$ is the sum of the external moments acting on a given human body segment,

\mathbf{I}^{cm} is the inertia matrix with respect to the center of mass,

$\boldsymbol{\omega}$ is the angular velocity vector,

$\dot{\boldsymbol{\omega}}$ is the angular acceleration vector.

The summations on the left hand side of these equations include terms due to gravity, external forces, and intersegmental forces and moments. The intersegmental forces and moments are generally solved recursively (Winter 2009).

Since intersegmental forces and moments are derived from these rigid body equations, their computation requires the estimation of segment mass, the position of the center of mass (CoM), and its inertia tensor (moments and products of inertia). All of these quantities must be transformed into a common coordinate system prior to estimation of the intersegmental forces and moments. Body segment inertial parameters (BSIPs) can be obtained using regression equations based on subject's segment length and body mass (De Leva, 1996a), geometrical models (Yeadon, 1990; Hanavan, 1964), optimization of parameters via COP errors in various postures (Chen, et al. 2011), or, when available, directly from subject-specific medical imaging (Ganley et al., 2004; Mungiole et al., 1999; Cheng et al., 2000). These estimations of BSIPs are the basis for rigid body models. The assumption that the estimates are constant are a source of uncertainty in model output.

For most regression equations, the position of the segment center of mass is given with respect to the proximal and distal endpoints, which define the segment length (De Leva, 1996a) or with respect to a number of anatomical landmarks (Zatsiorsky et al., 1990). Consequently, the determination of the BSIPs also depend on estimation of the joint centers positions. Adjusted scaling equations have been proposed for computing 3D inverse dynamics in which BSIPs are expressed in standardized definitions of the relevant anatomical axes (Dumas et al., 2007, Wu et al., 2002, Wu et al., 2005).

Due to specific anthropometric characteristics, BSIPs estimated in different populations can lead to different values (Nguyen et al., 2014). Uncertainties in the identification of BSIPs can play a critical role in reliable joint moment estimation, especially when analyzing motor activities involving high accelerations, such as in running (Krabbe et al., 1997) or when the population under examination has special anthropometric features, such as amputees (Sawers and

Hahn, 2010). To the contrary, when analyzing level walking at natural speeds, a minor influence of uncertainties in BSIPs is expected, particularly at the more distal joints in the stance phase (Rao et al., 2006; Camomilla et al., 2017).

Summary and Recommendations

The anthropometric model used to estimate body segment parameters must be detailed in order for results to be replicated. This includes procedures for estimating moments of inertia, mass, and center of mass locations. The sample for which regression equations were established should be consistent with the subjects being studied. This becomes especially important as linear and angular accelerations increase and for specific populations that may have substantially different BSP's (e.g. children, amputees...).

Joint centers

To compute the intersegmental joint moments, a reduction point, that is the point with respect to which the system of forces is reduced, is required. This point is classically defined as a joint center. In most of the human movement analysis protocols proposed in the literature, adjacent bony segments are conceptually assumed to be connected by spherical pairs, and their relative motion is described by three joint angles about the three anatomical axes defining the joint coordinate system and passing through this joint center (Wu et al. 2002, 2005). Then, when the joint allows only a small rotation about one axis (resisted degrees of freedom (DoF), e.g. adduction-abduction at the knee or elbow joint), it can be assumed, in a first approximation, that the relevant moment represents the action of the main anatomical joint restraints (i.e. articular contacts and ligaments) (O'Connor et al., 1998). On the other hand, in case of large rotations about one joint axis and neglecting friction (unresisted DoF, e.g., flexion-extension, adduction-

abduction and internal-external rotation at the hip or shoulder joints), the resultant of articular contact forces passes through this joint center and therefore it can be assumed that the relevant moment represent the actions of the muscle-tendon units (Challis et al., 1996). To interpret the joint moments with this rationale, it is required that the origin of the joint anatomical axes coincides with the reduction point (joint center) and that the joint moments are expressed then about these joint axes.

Joint centers are commonly defined by using regression equations from palpated external anatomical landmarks (Bell et al., 1990), using functional approaches (Leardini et al., 1999), multi-body kinematic optimization techniques (Begon et al., 2018) or using medical imaging techniques (Della Croce et al., 2005). The latter approach, although very accurate and able to create individualized musculoskeletal models, requires access to expensive and cumbersome measurement systems, time-consuming post-processing, highly-specific and multidisciplinary expertise, and in some cases, involves exposure to ionizing radiation. This explains why medical imaging techniques are frequently used as gold standards in biomechanics for the development and validation of models and techniques, but their use is limited in the fields of clinical movement analysis and sport applications.

It has been largely demonstrated that joint moment estimates are very sensitive to errors in the joint center and that this inaccuracy affects the calculations to a larger extent than other concurring factors such as errors on BSIPs (Camomilla et al., 2017). For instance, a hip joint anterior and lateral mislocation of 30 mm, which can be expected in typical of routine gait analyses (Hara et al., 2016), can cause a mean error of about 1.43 and 1.38% bodyweight x height in the flexion–extension and abduction–adduction moment components of the corresponding range, respectively (Stagni et al., 2000). In general, the strengths of the functional

method are that the calculated center of rotation is subject specific, side specific i.e. right different from left, and not influenced by the presence of bony or joint deformity. Nor is it influenced by differences in body segment proportions, as expected for gender, age, genetic traits, etc. On the other hand, its implementation requires the subject to perform, either passively or actively, a sufficiently wide joint angular excursion. Conversely, regression methods can be applied when movement restrictions are present. However, their accuracy and repeatability strongly depend on the original regression model, and are affected by uncertainties in anatomical landmark identification (Sangeux, 2015). The hip joint center is certainly the most critical lower extremity joint due to the large distance from the palpable pelvic landmark. It has been determined that the functional approach has errors between 10 to 20 mm whereas regression methods find errors between 15 to 30 mm (Leardini et al., 1999; Cereatti et al., 2009; Sangeux et al., 2014; Harrington et al., 2007).

Summary and Recommendations

Since joint center positions are used to define the moment arm of the forces acting on the segment under analysis, the manner in which they are identified will influence the estimation of intersegmental moments. Furthermore, because joint centers are commonly used to define segment length, they also affect body segment inertial parameters. It is therefore fundamental to use valid methods, and to clearly state these methods, for joint centers determination.

Signal Processing

Correct application and complete reporting of signal processing methods are crucial when dealing with kinematic and kinetic data. Of first concern is the sampling, which must be of an adequate rate to insure the frequencies present in the motion are completely captured. At a

minimum, the sampling rate must be greater than twice the highest frequency in the signal. A sampling frequency below this threshold will not only miss higher frequencies, the higher frequencies will fold back into the data and result in a contaminated signal (Edwards et al., 2017). This minimum sampling rate insures no information is lost but if peak values need to be accurately digitized the signal must be sampled at a much higher rate (5 to 10 times the highest frequency in the signal) or the digitized signal must be reconstructed using resampling techniques (Hamill et al., 1997). In general, movements that contain collisions are composed of higher frequencies and therefore must be digitized at a higher rate.

Kinematic data must be differentiated to calculate velocities and again to calculate accelerations in preparation for use in the equations of inverse dynamics. This double differentiation process amplifies the time series recording by the square of the frequency (Antonsson and Mann, 1985). Thus, high frequency noise can dominate the acceleration signal if proper processing procedures are not utilized. Smoothing techniques attempt to attenuate frequencies that comprise the noise while leaving the true signal unaffected. Various methods such as splines, time domain filters and frequency domain filters have been used to accomplish this. Selection of the frequencies that are being attenuation may be done using a set value or algorithms that objectively identify the cutoff frequencies (Jackson, 1979; Giakas and Baltzopoulos, 1999).

There is a concern that the frequency content of kinematic and kinetic data should be in agreement. Several researchers (Bisseling and Hoff, 2006; Kristianslund et al., 2012; Edwards et al., 2011) have shown that a disparity in the frequency content can cause artifacts in the intersegmental moments that cannot be explained by the dynamics of the activity. This necessitates the same cutoff frequencies for kinetic and kinematic data filtering. However, this

poses a problem with impact forces that can be attenuated by low-pass cutoff frequencies that are in the range necessary to reduce noise in the kinematic data. Decisions concerning the cutoff frequency need to be made based on the purpose of the experiment and the variables being measured.

Summary and Recommendations

Both kinematic and kinetic sampling frequencies must be clearly identified. The method of smoothing should be identified and the degree of smoothing (typically in the form of the frequency response) should be noted. The technique used to differentiate the data and any specialized techniques such as optimized cutoffs, resampling of data and procedures to minimize artifact should be detailed and cited.

Method of Calculation

There are two equivalent methods to describe the dynamics of a mechanical system, namely the Newton-Euler and Lagrange formulations. In terms of interpretation, the differences between the two are generally procedural rather than substantive. Note that in biomechanics, few human joints involve translational DoF greater than a few millimeters, thus, most of the time, the Lagrange equations of motion only result in moments. The Newton-Euler method is simpler and gives access to the full three-dimensional intersegmental force and moment vectors, including the moments for resisted degrees of freedom, such as knee adduction-abduction (Winter, 2009). Lagrange methods are especially useful when joint models that are more complex than spherical or hinge joints are needed, but do not solve for loads associated with resisted degrees of freedom (van den Bogert et al., 2013). Equivalent moments result from using the Newton-Euler equations of motion projected onto the DoF axes (projection with a dot product). These moments about the

joint DoF are directly related to the joint power (they just need to be multiplied by the DoF angular velocity) and can be described as "motor" or internal joint moments. In musculoskeletal modelling, these are typically the moments involved in the computation of the musculotendon forces while the other components of the intersegmental moments are assumed to represent the actions of ligaments and contact forces (Delp et al., 2007). Although theoretically equivalent, these two methods may produce small deviations because of differences in soft tissue artifact propagation.

Both the Newton-Euler and the Lagrange equations (Eberhard, 2006) lead to inverse dynamics procedures, meaning the intersegmental forces and moments are derived from the kinematics. In forward dynamics procedures, a muscle-driven or torque-driven model is used to estimate intersegmental moments from a neural signal obtained via electromyography, optimization procedures, or a combination (Buchanan, 2005).

When inertial components are absent or negligible, a static analysis was used to roughly estimate intersegmental forces and moments (Fantozzi et al., 2012). This simplified method consists of multiplying the ground reaction force vector by its moment arm at each joint and has been described as the 'ground reaction technique'. This method assumes that segment accelerations and/or the body segment inertial parameters are negligible, large errors can arise if this assumption is not met (Wells, 1981).

Summary and Recommendations

In general, static analysis of the human body should be restricted to static or near-static situations. Newton-Euler and Lagrange formulations of intersegmental moments are mathematically equivalent but the method should be identified because their sensitivity to signal

processing methods can be different. Forward or inverse dynamics procedures also need to be specified. If muscle-driven forward dynamics are used additional methods detailing the estimation of muscle forces are necessary.

Coordinate System

Intersegmental forces and moments have been presented in a variety of coordinate systems: global (also known as inertial or laboratory), proximal, distal and the joint coordinate system (Schache and Baker, 2007). In general, presentation of intersegment forces and moments in the global coordinate system should be avoided. Unlike segment coordinate systems, the intersegmental forces and moments presented in the global coordinate system will be affected by changes in the direction of motion.

The joint coordinate system (JCS) is appealing if kinematics are also presented in the JCS, but caution should be used because the axes of the JCS are not orthogonal. Projection onto non-orthogonal axes is problematic when the moment norm is to be computed or when the 3D vector is to be retrieved. If a JCS is used, the projection using a dot product (as opposed to a non-orthogonal projection) is preferred (Kristianslund et al., 2014). Those projected moments can be multiplied by the rates of change in the JCS angles to obtain a mechanically consistent joint power analysis. Also these orthogonally projected moments obtained from the Newton-Euler method will be identical to moments obtained from the Lagrange method where the JCS mechanism is explicitly modeled.

Proximal or distal segment coordinate systems (Figure 2) are useful in answering particular research questions. For instance, during the estimation of tibial tissue stresses, the intersegmental forces and moments would be expressed in a proximal segment coordinate system

at the ankle or a distal segment coordinate system at the knee so they are in a coordinate system that is suitable for further analysis, e.g. with a finite element model (Derrick et al., 2016).

Segment coordinate systems (and the joint coordinate systems derived from them) should be defined using ISB standards (Wu et al., 2002, 2005). If non-standard coordinate systems are used, they should be fully specified in terms of anatomical landmarks or other suitable definitions.

The choice of a coordinate system used to report intersegmental forces and moments can dramatically affect the interpretation of data. Note that during walking (Figure 3) and running (Figure 4) the sagittal plane moments and the vertical forces are similar between proximal and distal coordinate systems but there are some relatively large differences in the other planes and axes.

The choice of intersegmental coordinate systems should be consistent with the kinematics and the anthropometric model. Consistency will take two forms: mathematical and informational. Mathematical consistency is necessary to prevent inaccuracies that result from calculations with variables in more than one coordinate system. For instance, if moments of inertia from the anthropometric model were calculated in a segment coordinate system they should not be multiple by angular accelerations in a global or joint coordinate system. Likewise, subsequent calculations using intersegment forces and moments such as joint powers and apparent joint stiffness must have a consistent coordinate system to be accurate. Care should also be taken when estimating muscle moment arms using a musculoskeletal model. The kinematics applied to the model should be in the same coordinate system as the intersegmental moments if they are to be used in a common calculation. Informational consistency suggests that all quantities presented in a paper should be in the same coordinate system unless there is a

justifiable reason. This will remove ambiguity and instil confidence in the reader or reviewer that proper procedures have been followed.

Summary and Recommendations

The choice of the coordinate system (global coordinate system, proximal segment coordinate system, distal segment coordinate system, or joint coordinate system) highly influences the intersegmental forces and moments. It is therefore essential that the coordinate system used to interpret the intersegmental forces and moments be carefully considered and reported. Much thought and debate has gone into standardizing kinematic coordinate systems (Wu et al., 2002/2005) and the motivation for using JCS's for intersegmental forces and moments holds. Unless a rationale exists these previously defined kinematic coordinate systems should also be used to present intersegmental forces and moments. While describing coordinate systems, the directional signs of the forces should be defined (e.g., superior, anterior and lateral are positive) as well as for the moments (e.g., flexion, adduction, and internal rotation are positive) if these parameters are utilized. If moment signs are This is preferred over defining the x, y and z axes of the coordinate system because it gives additional information to the reader.

Internal or external perspective

Intersegmental forces and moments can be viewed from two perspectives. From an ***external point of view*** the intersegmental forces and moments represent the result of forces acting on the body as well as centrifugal and Coriolis actions arising from motion of the body segments. From an ***internal point of view*** these variables represent forces and moments that originate from within the body and act to resist external load and to maintain posture or accelerate the segments. Anatomical structures crossing the joint such as skin, fat, fascia,

muscles, ligaments, together with friction and contact between the articular surfaces produce the mechanical action. Both perspectives are equally valid and in fact the numerical value of the result is equal and opposite because the joint system is in balance. From the internal point of view an extension knee moment is present during mid-stance to prevent knee collapse but from an external point of view there exists a flexion knee moment caused by the external ground reaction force (passing posteriorly to the joint center in the static analysis perspective).

The decision to present internal or external intersegmental forces and moments is often made based on the researcher's view of the source of the moments. Internal moments are considered to be primarily caused by muscles when the joint is not near the end range of motion. This presents a problem at the knee joint because the result of all the muscles spanning the knee may produce a relatively small adduction-abduction moment, the primary source is considered to be ligaments and articular surfaces. This has led many researchers to present adduction-abduction moments at the knee using an external perspective when this is the primary variable of analysis (Telfer et al., 2017). This leads to additional confusion when the moments are referred to by their action. From an external perspective, an adduction moment at the knee is one in which the external forces are tending to cause the knee to adduct (ground reaction force passing medially to the joint center in the static analysis perspective), potentially tearing the lateral ligaments. However, from an internal perspective, an adduction moment at the knee is one in which the balance of muscles, ligaments and articular contact tend to adduct the knee. These differing perspectives can, and often do, lead to confusion for a reader trying to interpret knee function.

Summary and Recommendations

Whether intersegmental forces and moments are presented as internal or external can be determined by the research question being asked but may also be dependent on the perspective that the researcher is trying to convey. A clear statement of this perspective is essential to communicating concepts in the paper.

Normalization

In clinical movement analysis, demographic/anthropometric characteristics (i.e. age, height, body mass, gender) and the velocity ranges, with which the motor task is executed, influence the amplitude of the kinematic and kinetic variables and if not properly treated may act as confounding factors (Moisio et al, 2003; Senden et al., 2012; Andriacchi et al., 1977; Lelas et al. 2003). If these variables are not equivalent between the groups that are being compared they may need to be controlled statistically (covariate analysis) or their effect removed from the analysis (normalization). When a repeated measures study design is employed such that the normalization parameters do not change over time, the normalization process will not alter the statistical results. However normalization may still be warranted so that results can be conveniently compared to other studies.

Normalization procedures reduce the variance between individuals when comparing the intersegmental forces and moments among individuals. Most commonly, forces are divided by body weight or body mass, while moments are divided by these variables or quantities that result in non-dimensional values such as body weight multiplied by height or body weight multiplied by limb or foot length. For instance, before normalization, lower extremity peak moments during gait were statistically different between males and females in ten cases, but normalizing by body mass reduced this number to 6 and normalizing by body weight times height further reduced the number to 2 (Moisio et al., 2003). Pinzone et al. (2016) and Hof (1996) have also shown

advantages to using non-dimensional normalization. Additional reductions in residuals may be attained by considering more joint specific distance measurements or non-linear adjustments (Wannop et al., 2012, Lee et al., 2015).

Summary and Recommendations

Normalization of data is often necessary if groups are dissimilar on specific variables such as mass or height. This is especially useful in the context of gait laboratories in which individual data are frequently compared to a database. Although normalization will not change the statistical evaluation of a repeated measures study it may still be useful when comparing values to other studies. Normalization procedures need to be clearly outlined in the methods of the paper and normalization values such as average mass and height (or leg length) should be reported. Ranges of these values can also be useful so that researchers can avoid extrapolation of results. There is no simple method to normalize by the walking or running velocity therefore it is necessary to report average velocity or other variables that may influence the intersegmental forces and moments. This will assist researchers in explaining differences between studies.

Conclusions

Many options for the estimation and presentation of intersegmental forces and moments have been presented. These variations should not be considered correct or incorrect because each may be superior to the others in the context of the research paradigm and the questions being asked. However, as members of this field of research, we must insure that no ambiguity exists in the presentation of results. This is critical to an efficient evolution of a body of knowledge. We must be able to relate the curves and values presented by past researchers to the functional movement of the human body so that we can evaluate the results of the study, verify our own

research data, and ultimately create new theories and form new hypotheses. Realizing that there are differences in the detail of methodological information required in differing dissemination formats and in biomechanical vs clinical journals we make two suggestions: 1) even conference abstracts should include the coordinate system used and the internal/external perspective – interpretation of the results requires this minimum amount of information and 2) take advantage of the liberal policies journals have generally adopted that allow additional information to be posted online. In partial fulfillment of the goal of improved clarity in the scientific arena, we have identified three tangible items, in addition to this article, that we hope will help to fulfill this goal:

1. Reviewer/author checklist for presentation of intersegmental forces and moments. (Appendix A).
2. Online visualization tool for comparison of typical walking and running intersegmental forces and moments with adjustable coordinate systems, normalization methods and internal/external perspectives (ISB Website).
3. Software transparency. A request to the major biomechanics software companies to make easily available the items in the checklist so that users can access this information in a single location in the software.

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References (APA Style)

1. Andriacchi, T. P., Ogle, J. A., & Galante, J. O. (1977). Walking speed as a basis for normal and abnormal gait measurements. *Journal of biomechanics*, *10*(4), 261-268.
2. Antonsson, E. K., & Mann, R. W. (1985). The frequency content of gait. *Journal of biomechanics*, *18*(1), 39-47.
3. Baliunas, A. J., Hurwitz, D. E., Ryals, A. B., Karrar, A., Case, J. P., Block, J. A., & Andriacchi, T. P. (2002). Increased knee joint loads during walking are present in subjects with knee osteoarthritis. *Osteoarthritis and cartilage*, *10*(7), 573-579.
4. Begon, M., Andersen, M. S., & Dumas, R. (2018). Multibody kinematics optimization for the estimation of upper and lower limb human joint kinematics: a systematized methodological review. *Journal of biomechanical engineering*, *140*(3), 030801.
5. Bell, A. L., Brand, R. A., & Pedersen, D. R. (1987). Prediction of hip joint center location from external landmarks. *Journal of biomechanics*, *20*(9), 913.
6. Bell, A. L., Pedersen, D. R., & Brand, R. A. (1990). A comparison of the accuracy of several hip center location prediction methods. *Journal of biomechanics*, *23*(6), 617-621.
7. Bishop, C., Paul, G., & Thewlis, D. (2012). Recommendations for the reporting of foot and ankle models. *Journal of biomechanics*, *45*(13), 2185-2194.
8. Bisseling, R. W., & Hof, A. L. (2006). Handling of impact forces in inverse dynamics. *Journal of biomechanics*, *39*(13), 2438-2444.
9. Buchanan, T. S., Lloyd, D. G., Manal, K., & Besier, T. F. (2005). Estimation of muscle forces and joint moments using a forward-inverse dynamics model. *Medicine & Science in Sports & Exercise*, *37*(11), 1911-1916.
10. Burkhart, T. A., Andrews, D. M., & Dunning, C. E. (2013). Finite element modeling mesh quality, energy balance and validation methods: A review with recommendations associated with the modeling of bone tissue. *Journal of biomechanics*, *46*(9), 1477-1488.
11. Camomilla, V., Cereatti, A., Cutti, A. G., Fantozzi, S., Stagni, R., & Vannozzi, G. (2017). Methodological factors affecting joint moments estimation in clinical gait analysis: a systematic review. *Biomedical engineering online*, *16*(1), 106.

12. Cappozzo, A. (1984). Gait analysis methodology. *Human Movement Science*, 3(1-2), 27-50.
13. Cereatti, A., Bonci, T., Akbarshahi, M., Aminian, K., Barré, A., Begon, M., ... & Lin, C. C. (2017). Standardization proposal of soft tissue artefact description for data sharing in human motion measurements. *Journal of Biomechanics*, 62, 5-13.
14. Cereatti, A., Donati, M., Camomilla, V., Margheritini, F., & Cappozzo, A. (2009). Hip joint centre location: an ex vivo study. *Journal of biomechanics*, 42(7), 818-823.
15. Challis, J. H., & Kerwin, D. G. (1996). Quantification of the uncertainties in resultant joint moments computed in a dynamic activity. *Journal of Sports Sciences*, 14(3), 219-231.
16. Cheng, C. K., Chen, H. H., Chen, C. S., Lee, C. L., & Chen, C. Y. (2000). Segment inertial properties of Chinese adults determined from magnetic resonance imaging. *Clinical biomechanics*, 15(8), 559-566.
17. Chen, S. C., Hsieh, H. J., Lu, T. W., & Tseng, C. H. (2011). A method for estimating subject-specific body segment inertial parameters in human movement analysis. *Gait & posture*, 33(4), 695-700.
18. Chou, L. S., & Draganich, L. F. (1998). Increasing obstacle height and decreasing toe-obstacle distance affect the joint moments of the stance limb differently when stepping over an obstacle. *Gait & posture*, 8(3), 186-204.
19. Davis III, R. B., Ounpuu, S., Tyburski, D., & Gage, J. R. (1991). A gait analysis data collection and reduction technique. *Human movement science*, 10(5), 575-587.
20. De Leva, P. (1996a). Adjustments to Zatsiorsky-Seluyanov's segment inertia parameters. *Journal of biomechanics*, 29(9), 1223-1230.
21. De Leva, P. (1996b). Joint center longitudinal positions computed from a selected subset of Chandler's data. *Journal of biomechanics*, 29(9), 1231-1233.
22. Della Croce, U., Cappozzo, A., Kerrigan, D.C. (1999) Pelvis and lower limb anatomical landmark calibration precision and its propagation to bone geometry and joint angles. *Medical and Biological Engineering and Computing* 37(2), 155-161
23. Della Croce, U., Cappozzo, A., & Kerrigan, D. C. (1999). Pelvis and lower limb anatomical landmark calibration precision and its propagation to bone geometry and joint angles. *Medical & biological engineering & computing*, 37(2), 155-161.

24. Delp, S. L., Anderson, F. C., Arnold, A. S., Loan, P., Habib, A., John, C. T., ... & Thelen, D. G. (2007). OpenSim: open-source software to create and analyze dynamic simulations of movement. *IEEE transactions on biomedical engineering*, 54(11), 1940-1950.
25. Dempster, W. T. (1955). *Space requirements of the seated operator, geometrical, kinematic, and mechanical aspects of the body with special reference to the limbs*. Michigan State Univ East Lansing.
26. Derrick, T.R., Edwards, W.B., Fellin, R.E., & Seay, J.F. (2016) An integrative modeling approach for the efficient estimation of cross sectional tibial stresses during locomotion. *Journal of biomechanics* 49(3), 429-435.
27. Dillon, M. P., Barker, T. M., & Pettet, G. (2008). Effect of inaccuracies in anthropometric data and linked-segment inverse dynamic modeling on kinetics of gait in persons with partial foot amputation. *Journal of Rehabilitation Research & Development*, 45(9).
28. Dumas, R., Brånemark, R., & Frossard, L. (2017). Gait analysis of transfemoral amputees: errors in inverse dynamics are substantial and depend on prosthetic design. *IEEE transactions on neural systems and rehabilitation engineering*, 25(6), 679-685.
29. Dumas, R., Cheze, L., & Verriest, J. P. (2007). Adjustments to McConville et al. and Young et al. body segment inertial parameters. *Journal of biomechanics*, 40(3), 543-553.
30. Edwards, W. B., Derrick, T. R., & Hamill, J. (2017). Time Series Analysis in Biomechanics. In *Handbook of Human Motion* (pp. 1-24). Springer International Publishing.
31. Edwards, W. B., Troy, K. L., & Derrick, T. R. (2011). On the filtering of intersegmental loads during running. *Gait & posture*, 34(3), 435-438.
32. Eberhard, P., & Schiehlen, W. (2006). Computational dynamics of multibody systems: history, formalisms, and applications. *Journal of computational and nonlinear dynamics*, 1(1), 3-12.
33. Fantozzi, S., Garofalo, P., Cutti, A. G., & Stagni, R. (2012). 3D joint moments in transfemoral and transtibial amputees: When is the “Ground Reaction Vector

- Technique” an alternative to inverse dynamics?. *Journal of Mechanics in Medicine and Biology*, 12(04), 1250061.
34. Ganley, K. J., & Powers, C. M. (2004). Anthropometric parameters in children: a comparison of values obtained from dual energy x-ray absorptiometry and cadaver-based estimates. *Gait & posture*, 19(2), 133-140.
 35. Giakas, G., & Baltzopoulos, V. (1997). A comparison of automatic filtering techniques applied to biomechanical walking data. *Journal of Biomechanics*, 30(8), 847-850.
 36. Hamill, J., Caldwell, G. E., & Derrick, T. R. (1997). Reconstructing digital signals using Shannon's sampling theorem. *Journal of Applied Biomechanics*, 13(2), 226-238.
 37. Hanavan Jr, E. P. (1964). *A mathematical model of the human body* (No. AFIT-GA-PHYS-64-3). Air Force Aerospace Medical Research Lab Wright-Patterson AFB OH.
 38. Hara, R., McGinley, J., Briggs, C., Baker, R., & Sangeux, M. (2016). Predicting the location of the hip joint centres, impact of age group and sex. *Scientific reports*, 6, 37707.
 39. Harrington, M. E., Zavatsky, A. B., Lawson, S. E. M., Yuan, Z., & Theologis, T. N. (2007). Prediction of the hip joint centre in adults, children, and patients with cerebral palsy based on magnetic resonance imaging. *Journal of biomechanics*, 40(3), 595-602.
 40. Hof, A. L. (1996). Scaling gait data to body size. *Gait & posture*, 3(4), 222-223.
 41. Jackson, K. M. (1979). Fitting of mathematical functions to biomechanical data. *IEEE transactions on biomedical engineering*, (2), 122-124.
 42. Holden, J. P., & Stanhope, S. J. (1998). The effect of variation in knee center location estimates on net knee joint moments. *Gait & posture*, 7(1), 1-6.
 43. Kadaba, M. P., Ramakrishnan, H. K., Wootten, M. E., Gaine, J., Gorton, G., & Cochran, G. V. B. (1989). Repeatability of kinematic, kinetic, and electromyographic data in normal adult gait. *Journal of Orthopaedic Research*, 7(6), 849-860.
 44. Kirtley, C., Whittle, M. W., & Jefferson, R. J. (1985). Influence of walking speed on gait parameters. *Journal of biomedical engineering*, 7(4), 282-288.
 45. Krabbe, B., Farkas, R., & Baumann, W. (1997). Influence of inertia on intersegment moments of the lower extremity joints. *Journal of biomechanics*, 30(5), 517-519.

46. Kristianslund, E., Krosshaug, T., & van den Bogert, A. J. (2012). Effect of low pass filtering on joint moments from inverse dynamics: implications for injury prevention. *Journal of biomechanics*, 45(4), 666-671.
47. Kristianslund E., Krosshaug T., Mok K.M., McLean S., & van den Bogert A.J. (2014). Expressing the joint moments of drop jumps and sidestep cutting in different reference frames – does it matter? *Journal of biomechanics*, 47(1), 193-199.
48. Leardini, A., Cappozzo, A., Catani, F., Toksvig-Larsen, S., Petitto, A., Sforza, V., Cassanelli, G., Giannini, S., 1999. Validation of a functional method for the estimation of hip joint centre location. *Journal of Biomechanics* 32, 99–103.
49. Lee, D., Oh, S. E., Lee, I. K., Sim, T., Joo, S. B., Park, H. J., & Mun, J. H. (2015). Comparison of three normalization methods for 3D joint moment in the asymmetric rotational human movements in golf swing analysis. *Journal of Biosystems Engineering*, 40(3), 289-295.
50. Lelas, J. L., Merriman, G. J., Riley, P. O., & Kerrigan, D. C. (2003). Predicting peak kinematic and kinetic parameters from gait speed. *Gait & posture*, 17(2), 106-112.
51. Moision, K. C., Sumner, D. R., Shott, S., & Hurwitz, D. E. (2003). Normalization of joint moments during gait: a comparison of two techniques. *Journal of biomechanics*, 36(4), 599-603.
52. Mungiole, M., & Martin, P. E. (1990). Estimating segment inertial properties: comparison of magnetic resonance imaging with existing methods. *Journal of biomechanics*, 23(10), 1039-1046.
53. Nguyen, T. C., & Reynolds, K. J. (2014). The effect of variability in body segment parameters on joint moment using Monte Carlo simulations. *Gait & posture*, 39(1), 346-353.
54. O'Connor, J.J., Lu, T.W., Wilson, D.R., Feikes, J., & Leardini, A. (1998). Review: diarthrodial joints-kinematic pairs, mechanisms or flexible structures? *Computer Methods in Biomechanics and Biomedical Engineering*, 1, 123–150.
55. Pinzone, O., Schwartz, M. H., & Baker, R. (2016). Comprehensive non-dimensional normalization of gait data. *Gait & posture*, 44, 68-73.

56. PubMed search: “(biomechanics(tw) OR motion(tw) OR kinematics(tw)) AND recommendations(ti)”
57. Rao, G., Amarantini, D., Berton, E., & Favier, D. (2006). Influence of body segments’ parameters estimation models on inverse dynamics solutions during gait. *Journal of biomechanics*, 39(8), 1531-1536.
58. Sangeux, M. (2015). On the implementation of predictive methods to locate the hip joint centres. *Gait & posture*, 42(3), 402-405.
59. Sangeux, M., Pillet, H., & Skalli, W. (2014). Which method of hip joint centre localisation should be used in gait analysis? *Gait & posture*, 40(1), 20-25.
60. Sartori, M., Reggiani, M., Farina, D., & Lloyd, D. G. (2012). EMG-driven forward-dynamic estimation of muscle force and joint moment about multiple degrees of freedom in the human lower extremity. *PloS one*, 7(12), e52618.
61. Sawers, A. B., & Hahn, M. E. (2010). The potential for error with use of inverse dynamic calculations in gait analysis of individuals with lower limb loss: a review of model selection and assumptions. *JPO: Journal of Prosthetics and Orthotics*, 22(1), 56-61.
62. Schache, A. G., & Baker, R. (2007). On the expression of joint moments during gait. *Gait & posture*, 25(3), 440-452.
63. Scott, S. H., & Winter, D. A. (1990). Internal forces of chronic running injury sites. *Medicine and science in sports and exercise*, 22(3), 357-369.
64. Senden, R., Meijer, K., Heyligers, I. C., Savelberg, H. H. C. M., & Grimm, B. (2012). Importance of correcting for individual differences in the clinical diagnosis of gait disorders. *Physiotherapy*, 98(4), 320-324.
65. Stagni, R., Leardini, A., Cappozzo, A., Benedetti, M. G., & Cappello, A. (2000). Effects of hip joint centre mislocation on gait analysis results. *Journal of biomechanics*, 33(11), 1479-1487.
66. Telfer, S., Lange, M. J., & Sudduth, A. S. (2017). Factors influencing knee adduction moment measurement: a systematic review and meta-regression analysis. *Gait & posture*, 58, 333-339.

67. van den Bogert A.J., Geijtenbeek T., Even-Zohar O., Steenbrink F., & Hardin E.C. (2013). A real-time system for biomechanical analysis of human movement and muscle function. *Medical and Biological Engineering and Computing* 51, 1069-1077.
68. Wahid, F., Begg, R., McClelland, J. A., Webster, K. E., Halgamuge, S., & Ackland, D. C. (2016). A multiple regression normalization approach to evaluation of gait in total knee arthroplasty patients. *Clinical Biomechanics*, 32, 92-101.
69. Wannop, J. W., Worobets, J. T., & Stefanyshyn, D. J. (2012). Normalization of ground reaction forces, joint moments, and free moments in human locomotion. *Journal of applied biomechanics*, 28(6), 665-676.
70. Wells, R. P. (1981). The projection of the ground reaction force as a predictor of internal joint moments. *Bulletin of prosthetics research*, 10, 15-19.
71. Winter, D. A. (1984). Kinematic and kinetic patterns in human gait: variability and compensating effects. *Human movement science*, 3(1-2), 51-76.
72. Winter, D.A. (2009) *Biomechanics and Motor Control of Human Movement*, 4th Edition, Wiley & Sons.
73. Wu, G., Siegler, S., Allard, P., Kirtley, C., Leardini, A., Rosenbaum, D., ... & Schmid, O. (2002). ISB recommendation on definitions of joint coordinate system of various joints for the reporting of human joint motion—part I: ankle, hip, and spine. *Journal of biomechanics*, 35(4), 543-548.
74. Wu, G., Van der Helm, F. C., Veeger, H. D., Makhsous, M., Van Roy, P., Anglin, C., ... & Werner, F. W. (2005). ISB recommendation on definitions of joint coordinate systems of various joints for the reporting of human joint motion—Part II: shoulder, elbow, wrist and hand. *Journal of biomechanics*, 38(5), 981-992.

Conflict of Interest Statement

All authors were fully involved in the study and preparation of the manuscript.

The material within has not been and will not be submitted for publication elsewhere. We declare that we have no financial or personal relationships with other people or organizations that could inappropriately influence our work.

Appendix A. Example checklist for the reporting of intersegmental forces and moments.

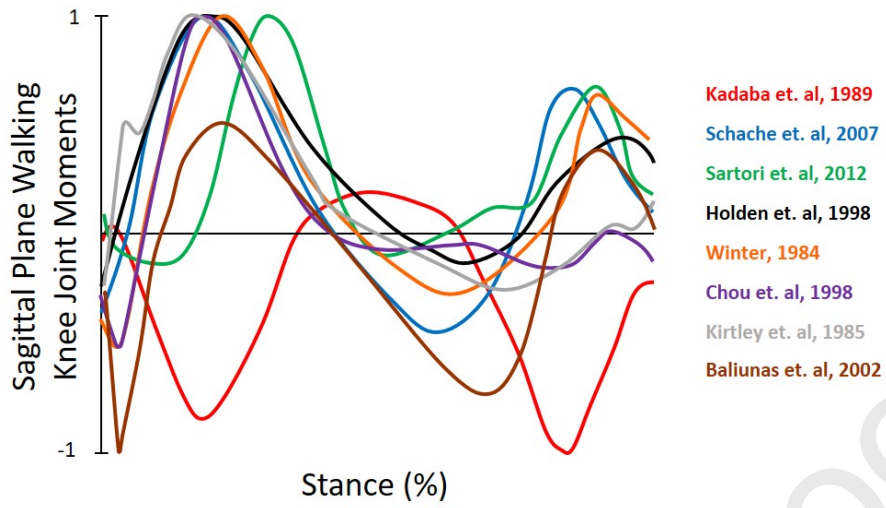
Anthropometric Model		Signal Processing	
Segment Masses	Dempster, 1955	Kinematics	Low-pass Filter @ 6Hz
Moments of Inertia	Hanavan, 1964	Kinetics	Low-pass Filter @ 6Hz
Joint Center Locations	Hip: Bell, 1989	Differentiation	First Central Difference Method
COM Locations	Hanavan, 1964	Other	
Method of Calculation		Internal or External Perspective	
<input checked="" type="checkbox"/> Newton-Euler	<input checked="" type="checkbox"/> Inverse Dynamics	<input checked="" type="checkbox"/> Internal	
<input type="checkbox"/> Lagrange	<input type="checkbox"/> Forward Dynamics	<input type="checkbox"/> External	
Normalization			
Moments		Forces	
	Average Value/Range		Average Value/Range
<input type="checkbox"/> Body Mass		<input type="checkbox"/> Body Mass	
<input checked="" type="checkbox"/> Body Weight	621 (449-721) N	<input checked="" type="checkbox"/> Body Weight	same as moments
<input type="checkbox"/> Body Mass x Body Height		<input type="checkbox"/> Other	
<input type="checkbox"/> Body Weight x Leg		<input type="checkbox"/> None	
<input type="checkbox"/> Other			
<input type="checkbox"/> None			
Coordinate System		Other	
<input type="checkbox"/> Global (Inertial)		Speed of Locomotion	3.5 m/s
<input checked="" type="checkbox"/> Proximal Segment		<input checked="" type="checkbox"/> Is the kinetic coordinate system the same as anthropometrics?	
<input type="checkbox"/> Distal Segment		<input checked="" type="checkbox"/> Is the kinetic coordinate system the same as kinematics?	
<input type="checkbox"/> Joint Coordinate System			
<input type="checkbox"/> Other			
Positive Force Directions			
<input checked="" type="checkbox"/> Proximal	<input checked="" type="checkbox"/> Lateral	<input checked="" type="checkbox"/> Anterior	
<input type="checkbox"/> Distal	<input type="checkbox"/> Medial	<input type="checkbox"/> Posterior	
Positive Moment Directions			
<input type="checkbox"/> Flexion	<input type="checkbox"/> Adduction	<input type="checkbox"/> Internal Rotation	
<input type="checkbox"/> Extension	<input type="checkbox"/> Abduction	<input type="checkbox"/> External Rotation	
or			
<input checked="" type="checkbox"/> Positive moment directions are defined by right hand rule of coordinate system			

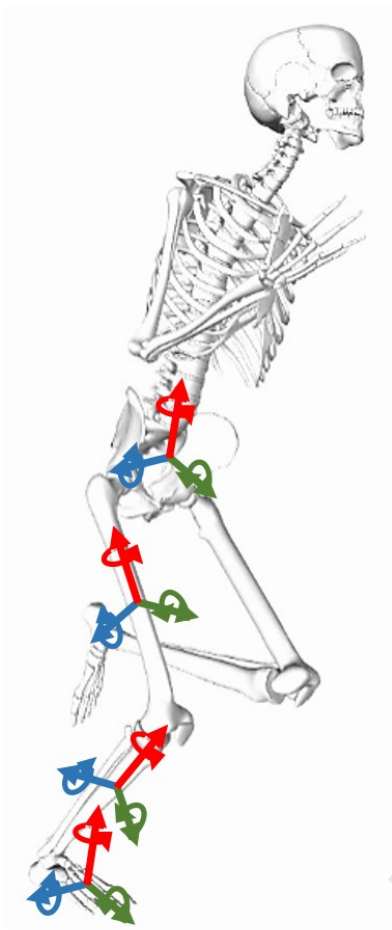
Figure 1. A selection of sagittal plane walking knee joint moments. Each curve is normalized to its own maximum absolute value. Various coordinate systems and methods of calculation result in an assortment of curve shapes. Average citations for these research papers is 478 (Google Scholar, May, 2019).

Figure 2. Segment coordinate systems for the pelvis, thigh, leg and foot segments. For the right leg these coordinate systems are defined by positive values pointing anterior, proximal, and lateral. Positive moments are clockwise about the axis while looking in the positive direction (right hand rule).

Figure 3. Ensemble averages of intersegmental forces (top) and moments (bottom) of eight subjects and five trials of walking (1.3 m/s). Both proximal (red) and distal (blue) coordinate systems are represented. Filled area represents ± 1 standard deviation of the proximal coordinate system. Kinematic and kinetic data were low-pass filtered at 6 Hz. Segment masses estimated using Dempster (1955). Segment moments of inertia and center of mass locations estimated using Hanavan (1964). Hip joint center estimated using Bell, Brand and Pederson (1989). Forces and moments were estimated using inverse dynamics with the Newton-Euler equations.

Figure 4. Ensemble averages of intersegmental forces (top) and moments (bottom) of eight subjects and five trials of running (3.5 m/s). Both proximal (red) and distal (blue) coordinate systems are represented. Filled area represents ± 1 standard deviation of the proximal coordinate system. Running kinematic and kinetic data were low-pass filtered at 10 Hz. Segment masses estimated using Dempster (1955). Segment moments of inertia and center of mass locations estimated using Hanavan (1964). Hip joint center estimated using Bell, Brand and Pederson (1989). Forces and moments were estimated using inverse dynamics with the Newton-Euler equations.





Journal Pre-proofs

