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HANDLING OF IMPACT FORCES IN INVERSE DYNAMICS


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Abstract

In the standard inverse dynamic method, joint moments are assessed from ground reaction force data and position data, where segmental accelerations are calculated by numerical differentiation of position data after low-pass filtering. This method falls short in analyzing the impact phase, e.g. landing after a jump, by underestimating the contribution of the segmental accelerations to the joint moment assessment.

This study tried to improve the inverse dynamics method for the assessment of knee moment by evaluating different cut-off frequencies in low-pass filtering of position data on the calculation of knee moment. Next to this, the effect of an inclusion of direct measurement of segmental acceleration using accelerometers to the inverse dynamics was evaluated.

Evidence was obtained that during impact, the contribution of the ground reaction force to the sagittal knee moment was neutralized by the moments generated by very high segmental accelerations. Because the accelerometer-based method did not result in the expected improvement of the knee moment assessment during activities with high impacts, it is proposed to filter the ground reaction force with the same cut-off frequency as the calculated accelerations. When this precaution is not taken, the impact peaks in the moments can be considered as artifacts.

On the basis of these findings, we recommend in the search to biomechanical explanations of chronic overuse injuries, like jumper's knee, not to consider the relation with impact peak force and impact peak moment.
Nomenclature

\[ \alpha_i = \text{angular acceleration of segment } i \]
\[ \omega_i = \text{angular velocity of segment } i \]
\[ a_i = \text{acceleration of the center of mass of segment } i \]
\[ F_e = \text{ground reaction force} \]
\[ g = \text{acceleration due to gravity} \]
\[ I_i = \text{moment of inertia of segment } i \]
\[ M_e = \text{external moment exerted on the foot} \]
\[ m_i = \text{mass of segment } i \]
\[ r_e = \text{position of application of ground reaction force} \]
\[ r_i = \text{position of center of mass of segment } i \]
\[ r_{jc} = \text{position of joint center (knee)} \]
\[ r_{pi} = \text{distance between attachment point } p \text{ of accelerometer and center of mass of segment } i \]
\[ g_{S_i} = \text{acceleration of the center of mass, including gravity, in Global Coordinate System} \]
\[ a_{S_{pi}} = \text{acceleration at attachment point } p \text{ of accelerometer of segment } i \text{ in local reference frame} \]
\[ g_{S_{pi}} = \text{acceleration at attachment point } p \text{ of segment } i \text{ in GCS} \]
2.1. Introduction

During impacts in activities such as running and jumping, impact force produces a shock wave which travels through the subject's body (Dickinson et al., 1985; Wakeling et al., 2003). Typical time histories of vertical ground reaction forces during impact show a high-frequency peak between 0 and 30 ms after touch down. A relation between this impact and chronic sports injuries has been suggested for many years (Collins & Whittle, 1989; James et al., 2003). Recent research, however, tends to shed doubts on a causal relation between impact forces and chronic overuse injuries. Bobbert et al. (1992) implied that the only way to control segmental rotations during impact in running were initial segmental kinematics and muscular activation levels prior to impact. Nigg (1986) stated that during impact, a person is not able to react with a change in muscle activity. It thus seems that initial kinematic conditions are primarily responsible for variations in impact forces, and that muscular activity does not play a major role. So, where joint moments as output of inverse dynamics represent active moments generated by the muscles, it can be concluded that impact force does not play a major role in the estimation of joint moments during impact and therefore cannot be related to chronic sports injuries (Gruber et al., 1998; Nigg & Wakeling, 2001).

Inverse dynamics analysis is a standard tool widely used for biomechanical studies. The standard inverse dynamic method (SM) to calculate joint moments uses a direct measurement of the external ground reaction force \( F_e \) and position data, while segmental acceleration is calculated by numerical differentiation of position data. With estimates of mass and inertial properties, joint moments can be calculated. The disadvantage of this procedure in estimating segmental acceleration is that differentiation of position data amplifies errors. These can partly be removed by filtering, but filtering has two outcomes. The first is a decrease in noise, but the second is removal of signal fluctuations faster than the cut-off frequency, which can distort the original signal contents. So, higher cut-off frequencies will result in less distortion of the original signal contents but will show an increase in signal noise. Subject to accurate position registration devices and high sample rates, better results in assessment of joint moments during fast transients like the impact phase can be obtained by using high cut-off frequencies.

Besides applying high cut-off frequencies to the raw position data for calculating segmental acceleration, another potential improvement is the use of an accelerometer-based method (AM), using accelerometers to measure the segmental acceleration accurately. In their study, using an experimental setup based on an instrumented compound pendulum, Ladin and Wu (1991) concluded that skin-mounted accelerometers might provide a viable, noninvasive approach that should be able to accurately estimate the acceleration of the underlying bone up to frequencies of hundreds of Hz, provided that the accelerometer is properly preloaded to the skin, to minimize errors due to soft tissue movement. For physical activities like running or jumping, this method is claimed to be much more accurate for estimating joint forces and moments than SM. An accurate estimation of joint forces and moments can give new insights or confirmation about the role of muscular activity during impact.

Direct measurement of acceleration has the disadvantage that accelerometers as a rule cannot be placed at the segment's center of mass. To find the center of mass acceleration, a correction is
necessary. Furthermore, there are practical difficulties to attach an accelerometer sufficiently rigid to the segment.

In the experiments to be described, we have tried to improve the assessment of the knee moment during the impact phase of landing after a jump by evaluating the effect of different cut-off frequencies in low-pass filtering on the calculation of knee moment. Next to this, we evaluated the effect of including direct measurement of segmental acceleration by accelerometer data in inverse dynamics.

2.2. Methods

Participants
Seven healthy well-trained male volleyball players participated in this study. Characteristics of the group of participants (mean±standard deviation) were: age 24±3 yr, body mass 79.49±8.17 kg, height 190.0±3.5 cm. The exclusion criteria were injuries at the lower extremities or the back in the previous 3 months. All participants gave their informed consent. The study was approved by the local ethics committee. The participants followed a standardized warming-up and stretching period. During the measurements, participants wore their own indoor sport shoes.

Procedures
Measurements were made after giving specific instructions to the subject. They were asked to perform a maximal countermovement jump. The participants were allowed to practice. Data acquisition was continued till five successful trails, with adequate landing on the force plate, were available for further analysis.

Data acquisition
Three-dimensional position data were collected at 200 Hz using an Optotrak motion analysis system with two cameras containing three sensors each. Two molded rigid marker frames (3.2 mm Aquaplastic), on which four light-emitting markers had been fixed, were tightly attached to the right thigh and shank with wide neoprene bandages, Velcro fasteners and adhesive tape. For the foot segment, four markers were attached to the shoe at the lateral side of the calcaneus.

Three components of the Fe, position of the center of pressure and three components of the external moment were recorded using a force platform (Bertec, type 4060-08).

For the AM, accelerometer data for shank and foot were collected using bi-axial accelerometers (Analog Devices, type ADXL150, sensitivity 38 mV/g, range±50 g). Both force-plate and accelerometer signals were sampled at 1000 Hz. After amplifying, all analog signals were converted to digital signals using the 16 bit A/D converter of the Optotrak system.

To collect 3D accelerometer data, two bi-axial accelerometers were mounted in a 26×26×28 mm box perpendicular to each other (weight 28 g). For the shank, the accelerometer unit was attached to the marker frame. For the foot segment, a special rigid frame was made of Aquaplastic (3.2 mm) to fasten the
Figure 2.1. A subject's configuration for the foot and the shank. The shank accelerometer was attached to the marker frame. The foot accelerometer was attached to a frame, fastened by tightening the shoelace. Foot markers were glued on the shoe.
accelerometer. This frame was placed on the tongue of the shoe and fastened by tightening the shoelace (Figure 2.1).

To characterize the position and orientation of the underlying bone of the three segments describing the anatomical coordinate system, an anatomical landmark calibration was carried out using an Optotrac 6-Marker probe. For this procedure, the subject was standing in an erect posture so that all markers of the marker frames were visible. To determine the position of the accelerometers with respect to the center of mass of shank and foot, both accelerometers were also pointed during the calibration procedure. The anatomical frame definition was according to Cappozzo et al. (1995). The global coordinate system (GCS) as well as the anatomical coordinate system were defined according the ISB recommendations (Wu and Cavanagh, 1995): the positive x-axis points forward, the positive y-axis points upward and the positive z-axis points to the right. The local knee joint coordinate systems (KJCS) was defined as follows: the origin was the midpoint between the lateral and medial femoral condyles, the z-axis coincident with the z-axis of the thigh coordinate system, the y-axis coincident with the y-axis of the shank coordinate system, and the x-axis perpendicular to y- and z-axes.

Anthropometric data for the estimation of segment mass, segment length, center of mass and radius of gyration for each plane of rotation of thigh, shank, and foot were calculated using data from de Leva (1996).

Data analysis

Three-dimensional dynamic analyses were done for the right limb using a 3-segment rigid body model. A Matlab (The Mathworks, Inc; version 6.5)-based motion analysis program BodyMech (Free University, Amsterdam) was used to process kinematic and kinetic data, and inverse dynamics was used to calculate 3D knee moments. Smoothing of analog data (both force-plate and accelerometer data) was done by a second-order low-pass Butterworth filter with a cut-off frequency of 100 Hz, applied in a zero-phase forward and reverse digital filter. The start of the landing phase of the countermovement jump was defined as the time when the vertical ground reaction force exceeded 4 N.

Standard method

For evaluating the effect of different cut-off frequencies on the calculation of knee moment in SM, kinematic data were smoothed at cut-off frequencies of 20 and 100 Hz (SM20 and SM100, respectively).

The calculation of knee moment $M_k$ in the GCS using position data and force plate data was based on the equation of motion formulated by Hof (1992):

$$M_k = -M_e \cdot (r_e \cdot r_{p_c}) \times F_e - \sum [(r_i \cdot r_{p_c}) \times m_i \cdot g] + \sum [(r_i \cdot r_{p_c}) \times m_i \cdot \dot{a}_i] + \sum \frac{d}{dt}(I_i \omega_i)$$

term1a term1b term2 term3 term4

For symbols see Nomenclature. The summations refer to the foot and shank segments.
Accelerometer Method

For the AM, accelerometers were used to measure accelerations instead of using the second derivative of position data. Using the AM, knee moments in GCS can be calculated as follows:

\[
M_k = -M_e - (r_e - r_J) \times F_e + \sum_i [(r_i - r_J) \times m_i g \dot{s}_i] + \sum \frac{d}{dt} (I_\omega)
\]

(2)

\[
\text{term1a} \quad \text{term1b} \quad \text{term2-3} \quad \text{term4}
\]

In Eq. (2) term 2 and 3 are combined because the accelerometers measure both linear acceleration \(\dot{a}_p\) and gravity \(g\), so the accelerometer signal at attachment point \(p\) in the local accelerometer coordinate system (ACS) is

\[
\dot{s}_p = \dot{a}_p - g
\]

(3)

In order to arrive at \(s_i\), the accelerometer output \(\dot{s}_{pi}\) should be transformed to the GCS. The relevant transformation was found by a method similar to the one used for determining anatomical reference frames from global reference frame kinematic data (Challis, 1995). The relevant transformation matrix relating the measured marker positions to the accelerometer orientation was assessed from a number of static postures with different segment orientations (thus different values of \(\dot{s}_p = g\)). To this point we have calculated the linear acceleration by accelerometers in the GCS. For the inverse dynamics we need the linear acceleration of the segment's center of mass. Accurate estimation of the linear acceleration of the centers of mass (Eq. (4)) was determined using a combination of direct measurements of segmental position, linear acceleration and angular velocity (Ladin & Wu, 1991). The relevant kinematic data were low-pass filtered at 100 Hz.

\[
\dot{s}_i = \dot{s}_{pi} + \alpha_i \times r_{pi} + \omega_i \times (\omega_i \times r_{pi})
\]

(4)

2.3. Results and discussion

To improve the assessment of the knee moment during the impact phase of landing after a jump, the SM using low-pass filtering at 20 Hz (SM20) will be compared with the SM100, low-pass filtered at 100 Hz, and with the AM, both accelerometer output and required kinematic data (Eq. (4)) filtered at 100 Hz. The results will be presented by means of the recordings of the sagittal knee moment in KJCS of a representative subject. All participants showed similar patterns in knee moments and segmental acceleration.
Standard method, filtered at 20 or 100 Hz: SM20 and SM100

From the curves in Figure 2.2 it can be seen that during impact, the knee moment calculated by SM20 showed the characteristic extension moment peak, whereas SM100 showed a gradual rise of the knee moment instead. SM100 showed during landing phase a rippled curve due to the higher cut-off frequency. During impact, the SM100 moment caused by the impact peak from the ground reaction force is largely counteracted by the moment caused by segmental acceleration (Figure 2.3). The curves of SM100 and SM20 in Figure 2.4 clearly show that during the impact phase, SM acceleration data are sensitive to the height of the cut-off frequency. SM100 showed much higher peak values than the accelerations filtered at 20 Hz. The real segmental accelerations thus contain frequencies well above 20 Hz. This feature demonstrates very clearly the underlying problem of SM20 in inverse dynamics: filtering accelerations at 20 Hz suppresses high-frequency components. So, when calculating segmental accelerations from these filtered position data, these accelerations cannot sufficiently counteract the impact peak given by the ground reaction force in the intersegmental moment calculation, which are commonly filtered at a higher frequency (at 100 Hz or not at all), resulting in a large but spurious extension impact peak by SM20.

![Figure 2.2. Comparison of the sagittal knee moments of landing that result from inverse dynamics using SM20 (black dotted line), SM100 (black solid line) and AM (gray solid line). A positive sagittal knee moment corresponds to an extension moment, and a negative sagittal knee moment corresponds to a flexion moment. Time scale starts at the first floor contact.](image)

Inverse dynamics with accelerometer data: AM

In most position registration systems, the noise level is such that kinematic data cannot be filtered at 100 Hz. Therefore, we tried with the AM to create an alternative to determine high-frequency components of the segmental acceleration more truthfully compared to SM20, which would result in a more realistic assessment of the knee moment, comparable to the results of SM100. Contrary to our expectations, the
Figure 2.3. Contributions of Fe (term 1b) and segmental accelerations (term 3 of SM and terms 2, 3 of AM, see Eqs. (1) and (2), to the sagittal knee moment (thick gray solid line). Thick black solid line is the contribution of term 1b. Acceleration contributions consist of contributions of the vertical (y, thin black dotted line) and horizontal (x, thin black solid line) accelerations of the foot, and the y (gray dotted line), and x (thin gray solid line) accelerations of the shank.

Figure 2.4. Comparison of the horizontal (x) and vertical (y) accelerations of the centers of mass of the foot and shank, determined from position data filtered at 20 Hz (SM20, thin black dotted line), at 100 Hz (SM100, thin black solid line), and from accelerometer data (AM, thick black solid line).
curves of the knee moment determined by AM showed a huge and above all a physiological unrealistic flexion moment peak (Figure 2.2). Similar to SM100, AM showed a rippled curve during the landing phase.

From the comparison of the different contributions to the knee moment of the different methods (Figure 2.3), one can conclude that this flexion moment peak in AM is mainly caused by the component related to the horizontal foot acceleration, as determined by the accelerometers. The only difference between SM100 and AM is the assessment of segmental acceleration. To get a closer look at the differences in segmental accelerations, Figure 2.4 shows the vertical and horizontal linear accelerations of the centers of mass of foot and shank during impact, derived from position data used in the SM20 and SM100, and from accelerometer data used in AM. Major differences in amplitude are particularly seen in horizontal foot acceleration, although impact acceleration of the SM100 and AM are calculated with the same cut-off frequency. The high horizontal foot accelerations of AM result in a very large contribution to the knee moment (terms 2, 3 in Eq. (2)), causing the unrealistic flexion peak.

**Model to verify the AM**

In order to verify the concept of AM assessing segmental acceleration, we used a simple mechanical model. This rigid one-segment model was used to compare the Fe and the force generated by linear acceleration of the segment, determined by accelerometers, during impact. Four cluster markers and one accelerometer unit were attached to a wooden beam (2.3 kg, length 35 cm) with a rubber cushion at the lower end. The same procedures were followed as mentioned in the methods section for determining linear acceleration in the GCS. To simulate a free fall of a body segment, the model was released from 30 cm above the ground. The curves in Figure 2.5 show an excellent correspondence between the measured Fe and the segment mass times linear acceleration, and therefore validate the accuracy of the AM.

However, in the practical situation of determining the acceleration of the human foot in landing, we have to admit that the overshoot in horizontal acceleration by the accelerometers in the AM remains unsatisfactory and unclear. At this stage, we have to conclude that this procedure does not result in the expected improvements in assessing the knee moment. In our search to find an explanation for the overshoot in accelerometer data, we filtered the position data used for the correction factor (Eq. (4)) at 20 Hz instead of the 100 Hz used in our method, but no improvements were found here. An explanation might be found in the rigid-body assumption of the foot. Mechanically, the human foot does not consist of a single rigid segment, but is quite a deformable structure. This implies that in reality during an impact, different accelerations of several segments within the foot contribute to the moment around the knee. Unfortunately, rigid attachment of two or even more accelerometers to separate foot segments gives practically insuperable technical problems. Although both AM and SM are applied to the same unrealistic rigid-foot model, the difference in the place of attachment and insufficient rigidity of attachment of the accelerometer unit may have contributed to the overshoot in measured acceleration.
Figure 2.5. Comparison of horizontal and vertical impact forces from Fe (solid line) and the force generated by linear accelerations, determined by AM of the mechanical model (dotted line).

Figure 2.6. Comparison of the sagittal knee moments of landing that result from inverse dynamics using SM100 (dotted line) and the proposed SM20-20 (solid line).
Proposal for alternative method

The curves in Figure 2.2 showed remarkable but expected differences between the inverse dynamical output of SM20 and SM100. For the reasons discussed above, we can conclude that the impact peak in the knee moment in SM20 is an artifact. The remaining question is, when studying impact dynamics, which inverse dynamical method gives the most truthful output. Somewhat to our regret, the use of accelerometer data cannot provide a complete solution to the inverse dynamical problem in this situation.

To assess knee moments in future work, without the incorrect impact peak, we suggest an adjustment in filter techniques of the analog Fe and position data in the SM. Because commonly used position registration systems are not suitable to use kinematic data filtered at a cut-off frequency of 100 Hz for the increase in noise, a method using a low cut-off frequency for filtering kinematic data is to be preferred.

In the broadly accepted SM20, we used for both kinematic and force plate data a second-order low-pass Butterworth filter, with a different cut-off frequency (20 and 100 Hz, respectively, applied in a zero-phase digital filter). To minimize the inaccuracy during impact and to calculate a more truthful knee moment we suggest to use the same cut-off frequencies and the same filter technique for both kinematic and force plate data. By filtering both signals at the cut-off frequency of the kinematic data, contribution of the impact peak force is sufficiently counteracted by the impact acceleration. The curves in Figure 2.6 show the comparison between this proposed inverse dynamical method with an equal cut-off frequency of 20 Hz (SM20-20) for both kinematic and force plate data, and the knee moment assessed by SM100.

Both the new equally filtered SM20-20 and SM100 clearly show a more gradual rise of knee moment during the impact phase, instead of a high transient impact peak, either an extension (SM20) or a flexion peak (AM). Such a gradual rise can be considered more realistic than the fast transients. Furthermore, the maximum moment of SM20-20 and SM100 around 80 ms corresponds well.

The choice of cut-off frequency is, as always, determined on the one hand by the desire to faithfully reproduce fast transients and on the other hand by the need to reduce noise in the accelerations. For the present setup, very precise Optotak position measurement at 200 Hz and for the present purpose, studying very fast movements in landing, a cut-off frequency of 20 Hz was considered sensible in view of the findings with SM100.

2.4. Conclusion

When estimating knee moments using the standard inverse dynamic method in activities like running and jumping, one should take into consideration the inaccuracy of the calculated accelerations during impact. We showed that the impact peak in Fe is the effect of very high segmental accelerations, which neutralize the contribution of Fe to the knee moment. To overcome inaccuracies in assessment of the knee moment, we suggest using the same cut-off frequencies and the same filter techniques for both kinematic and force plate data. On the basis of these findings, we recommend in the search to biomechanical explanations of chronic overuse injuries, like jumper’s knee, not to consider the relation with the impact peak moment.
References


