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Calculation of ankle and knee joint moments during ACL-injury situations in soccer

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Abstract

The basis of ACL-injury prevention is the understanding of the injury mechanism. Therefore a new approach was developed and validated that enables the calculation of knee and ankle joint moments during the injury. Detailed analysis of ACL-injury situations was performed to detect the kinematics as input data for a simplified 3D-human body model. An inverse-dynamics approach was used to realize the movement. The model was driven by Net-Muscle-Torque-Motors that calculate 3-D ankle and knee joint moments. Although there are some limitations that have to be considered this approach has the potential to generate a better understanding of injury mechanisms.

Keywords: ACL-injury; injury mechanism; computer simulation; joint torque; inverse dynamics;

1. Introduction

Injuries to the anterior cruciate ligament (ACL) are one of the most serious injuries in soccer [1]. Thus a lot of effort has been spent on the question how to prevent ACL-injuries, especially those that occur without any contact to an opponent player. Van Mechelen et al. [2] described a step by step process consisting of four parts that should be followed during the development of injury prevention measures. The basic requirement after identifying the problem is the understanding of the injury mechanisms, as the investigation is complex and mostly inexact and subjective.

Krosshaug et al. [3] collected eight different research approaches concerning analysis of injury mechanisms and discussed their strengths and weaknesses. They concluded that for most injury types no single research approach is adequate especially regarding the completeness of information provided. It is therefore necessary to combine a number of different approaches to describe the mechanisms completely. According to them the combination of video analysis and mathematical simulation could be one relevant approach having the ability to provide a broader and more precise understanding of the injury mechanism.

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Wallrapp et al. [4] stated that from the various methods applied to simulate human movement, forward dynamics and inverse dynamics are the most common ones. However joint loads of impacts obtained via inverse dynamics using models that inappropriately reflect the properties of the moving system have to be judged with caution [5]. More realistic joint forces can only be estimated using a forward dynamic model including real muscle force functions, acting at the real position and using correct force parameters. But such models are complex and mostly applied only to simple human motions [4]. However McLean et al. [6] described a sophisticated forward dynamic 3D musculoskeletal model of the lower extremity and applied it to a cutting movement.

Nevertheless the simulation approach offers the possibility to study different injury mechanisms in a computer environment avoiding any exposure to an athlete [3]. The problem in doing so is the need of kinematic data as input for the simulation model. Hence the investigation of "close to injury situations" is often the only way. But then the question is whether this situation really represents real injury patterns. Barone et al. [7] found in an accidental ACL-injury during a ski-jump landing analysis that the kinematics differed significantly between injury and non-injury situations.

Krosshaug and Bahr [8] introduced a new method providing a suitable opportunity to assess reasonable kinematic data from injury situations. Even though necessary input data could be assessed by this method, no attempts have been undertaken to calculate joint loads, so far.

Therefore the presented study had two main goals: Firstly, develop and validate an approach that enables the application of computer simulation on real ACL-injury situations in order to calculate knee and ankle joint moments. Secondly, apply the new method on real ACL-injury situations and calculate ankle and knee joint moments.

2. Methods

2.1. Analysis of ACL-injury situations

Analysis of non-contact ACL-injury situations in elite soccer was done by means of the Poser-Method described by Krosshaug and Bahr [8]. Three videos showing injuries from at least three different perspectives were selected for further analysis. Within the 3-D modelling software, Poser® 4 and Poser® Pro Pack (Curious Labs, Inc., Santa Cruz, CA, USA) a skeleton model (Zygote Media Group, Inc., Provo, UT, USA) was matched in the different camera views, frame by frame into the injured player [8], as shown in figure 1. Calibration is done by characteristic landmarks with standardized dimensions such as marking lines or the goal. The dimensions of the skeleton were fitted as exact as possible to the anthropometry of the injured player so that the body height of the model and the subject matched. The matching of every injury situation started during the flight phase at least 80 ms before initial ground contact of the injured leg and was continued until its complete unloading or until the collapse of the player after the injury.

The Poser data was loaded into Matlab version R2006a (The Mathworks, Inc., Natick, MA, USA) for post-processing. Joint angles, angular velocity and acceleration for all joints of the skeleton model, centre of mass (COM) for each segment and the whole model and velocity and acceleration of the COM were calculated. Furthermore the ground reaction force (GRF) occurring during ground contact was approximated applying Newtons second law, see equation (1)

$$F = m * a \qquad (1)$$

2.2. Simulation Model

A simplified 3-D rigid body model of the human body [4] based on the model described by Hanavan [9] was used for simulation, see Fig. 1. The model was built in the multi-body-simulation (MBS) tool SIMPACK (Simpack AG, Gilching, Germany) and consisted of 16 segments that were connected with spherical joints each with three degrees of freedom. The sub-division of the torso into three parts corresponds to the one of the Zygote-skeleton model used in the Poser analysis. As no exact anthropometric data of the injured players were available the

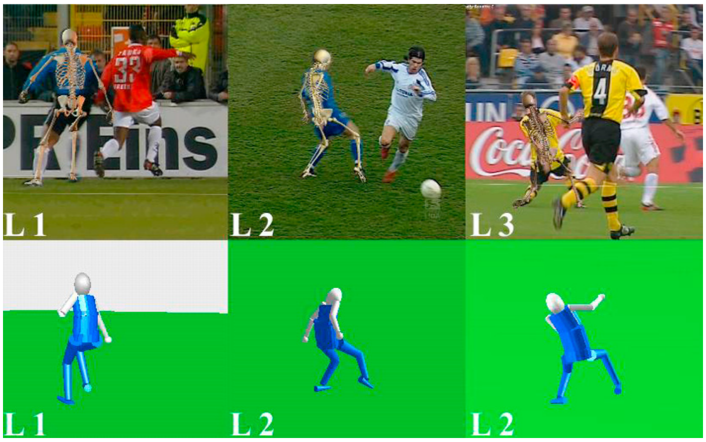

Fig. 1. Screenshots of the Poser analysis (top) and of the visualization of the human body model in SIMPACK (below).

necessary inputs were calculated using the method of Zatsiorsky [10]. Ankle and knee joints were driven by net Net-Muscle-Torque-Motors using a cascaded control law [4], see equation (2)

$$T_{mj} = K_v \left(K_p \left(\hat{\beta}_j(t) \right) - \beta_j \right) - \dot{\beta}_j \qquad (2)$$

where K_v and K_p are control parameters of the control law, $\hat{\beta}_j$ is the desired joint angle and β_j and $\dot{\beta}_j$ are actual angle and angular velocity, respectively. Additionally linear torsional damping was added to decrease noise. The same control parameters were used for all axes of the two joints. For all remaining joints of the human MBS standard rheonomic joints were used to drive the MBS-model. External loads were applied as functions of time via standard force elements in SIMPACK. After completion of the simulation the calculated joint moments were retransferred into Matlab and smoothed by a weighted average filter.

2.3. Model validation

Validation of the model was done in two parts. First, we wanted to figure out which factors influence inaccuracies of the Poser kinematics and the GRF approximation have on joint torques compared to more exact input data from motion analysis and force plate measurements. Therefore we used data from laboratory measurements of a plant-and-cut movement and the corresponding most exact available poser data [8]. Second, the correctness of the calculated joint moment in general was verified by comparing them with moments estimated by an iterative Newton-Euler approach [11] in Matlab. This validation was also done with the data from the laboratory measurements.

Furthermore RMS-values of the difference between desired and actual joint angles were calculated to secure that the simulation model executed the specified movement correctly.

2.4. Application on real injury situations

Kinematic data and the calculated GRF of the analyzed injuries were taken as input for the simulation process. In order to get not too noisy joint moment the input data was preliminarily smoothed by a weighted average filter in Matlab and then transferred onto the SIMPACK model. As well as for validation, RMS-values of the angular deviation between actual and desired joint angles were calculated.

3. Results

3.1. Kinematic data

Poser analysis delivered time histories for the joint angles. Exemplarily knee joint angles during ground contact of the injured leg are shown in Table 1. Analysis of the velocity of the body COM revealed that in each situation the injured athlete decelerated. Approximation of the GRF showed values up to 3400 N in vertical direction according 2,5 to 3,8 times body weight of the player depending on the injury situation, see Table 1.

Furthermore the horizontal components of the GRF necessary as input for the computer simulation were calculated by Equation 1. The values given exemplarily in Table 1 are the horizontal force components in anterior-posterior and medial-lateral direction, respectively that occur at the time of the maximal vertical GRF.

Table 1. Kinematic data derived from Poser analysis. Joint angles are given for the complete period of ground contact of the injured leg. (AP-GRF is the anterior-posterior component (anterior: +), ML-GRF is the medial-lateral component (medial: +)).

Injury Situation	Flexion (+) [°]	Valgus (+) [°]	Int. rot. (+) [°]	Max. vertical GRF [N]	player BW [g]	AP-GRF [N]	ML-GRF [N]
L 1	42 - 73	-7 - +8	-3 - +1	3396	3,8	-460	517
L 2	36 - 73	0 - 1	1 – 4	1653	2,5	-1288	1034
L 3	33 - 66	-4	0	2530	3,4	-1930	-1307

3.2. Validation

Comparison of the SIMPACK model and simple inverse dynamic calculation showed slightly different characteristics for the simulated movement. Within the first 100 ms after initial contact maximal joint moments were almost equal for flexion of knee and ankle and internal rotation of the ankle, respectively. Valgus moment (see Fig. 2) and internal rotation moment of the knee oscillated during the first 100 ms after impact. For both the net-muscle-torque model underestimated the first (higher) peak but matched the second peak with good accuracy. Only the adduction moment of the ankle joint was overestimated clearly in the SIMPACK model compared to inverse dynamics. With increasing time all joint moments of the knee joint and internal rotation moment of the ankle joint were overestimated, however the qualitative characteristic of the curves suited. In contrast to this ankle flexion and adduction matched well.

Input data seemed to have a considerable influence on the simulation results. The result curves from Poser data were shifted compared to the one calculated with motion analysis data. Qualitative characteristics of the graphs were very similar to the SIMPACK model using exact input data, see Fig. 2. The results from Poser data were noisier than the one from measurements. However maxima and/or minima of the curves differed only within a reasonable magnitude but occurred later compared to exact data simulations (ankle flexion moment earlier).

RMS-values for the lab-trial model lay between 0,1007° and 0,3707° for the ankle and between 0,0719° and 0,414° for the knee, respectively and indicate that the movement pattern of the simulation model was correct.

3.3. Ankle and Knee Joint Torque

Ankle and knee joint moments were calculated for the three ACL-injury situations, see Fig. 3. RMS-values indicating the accuracy of the simulated movement showed generally only small values between 0,022° and 1,32° over all simulations. The greatest deviations were found in the ankle joint (supination and internal rotation) during L 1, the smallest deviations for L 2.

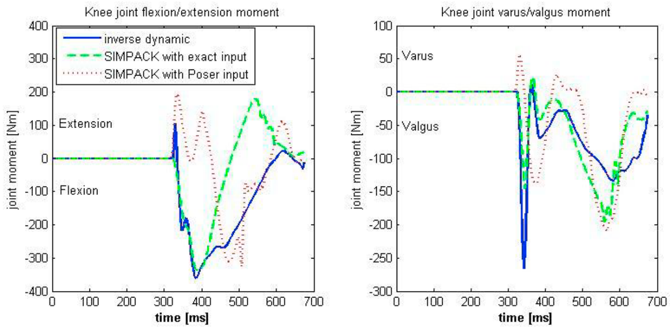

Fig. 2. Comparison of SIMPACK simulation and simple inverse dynamic calculation for two axes of the knee joint (initial ground contact was at t=320 ms)

4. Discussion

It is generally accepted that the use of computer simulation could provide a lot of additional information on the injury mechanisms as internal forces and moments are determined. Therefore a new approach of combining detailed ACL-injury analysis and computer simulation using net-muscle-torque-motors had been developed.

Krosshaug and Bahr stated that for a good quality matching at least three different camera perspectives are necessary preferably perpendicular to each other [8]. We met this demand as we analysed only injury situations where three (L 2 and L3) or four camera perspectives (L 1) were available. Thereof at least one camera showed the injured player in detail that enhanced the quality of the matching. However, high frequency oscillations shortly after impact could only be determined with limitations [8]. Furthermore the great distance between cameras and injured player resulting in a small size of the player within the images complicates the analysis and decreases accuracy. Nevertheless this method provides the possibility to obtain reliable kinematics from real injuries that don't occur during scientific measurements and it is much more exact than only visual video inspection or other methods.

Computer simulation was conducted using a simplified MBS-model of the human body in SIMPACK consisting of 16 rigid segments. By the use of net-muscle-torque-motors we were able to realize the kinematics obtained via Poser analysis with good accuracy and to calculate knee and ankle joint moments. However some limitations have to be considered. First, the human body model used here is a rigid body model. That means that the results have to be interpreted with great caution as no damping characteristics of the human body were considered, yet. According to Gruber et al. especially in the impact phase the moments could therefore be too high [5]. To meet this problem a damping element was implemented in the net-muscle-torque motors. Accordingly the joint moments of the injury situations did not show high impact peaks at or shortly after initial contact. Thus we assume that the results at least during 80ms to 150ms after impact should be reasonable.

Second, we used the GRF determined by the Poser method as external forces for the simulation. These forces were only approximations and possibly defective. Further we did not attempt to model the shoe-surface interaction, yet. The modeling of the interaction of studded shoes with natural grass is complex and depending on many factors like surface and its condition, sole material, stud design, -alignment and –material and others. Therefore we think that the approach of using approximated GRF data as input for the simulation is an adequate way to model injury situations that occurred during soccer games and not in the lab.

Third, the chosen inverse dynamic modelling approach is not capable to determine exact joint forces [4]. On the other hand forward dynamic models that do not include all muscles – at least of the lower extremity – and that do not represent the anatomic reality (bones, ligaments, menisci, cartilage, wobbling masses, etc.) in detail will not deliver absolutely correct results. Thus, to our opinion, the chosen modelling approach is a good compromise between complexity and benefit, especially as a first step towards a more complex computer simulation.

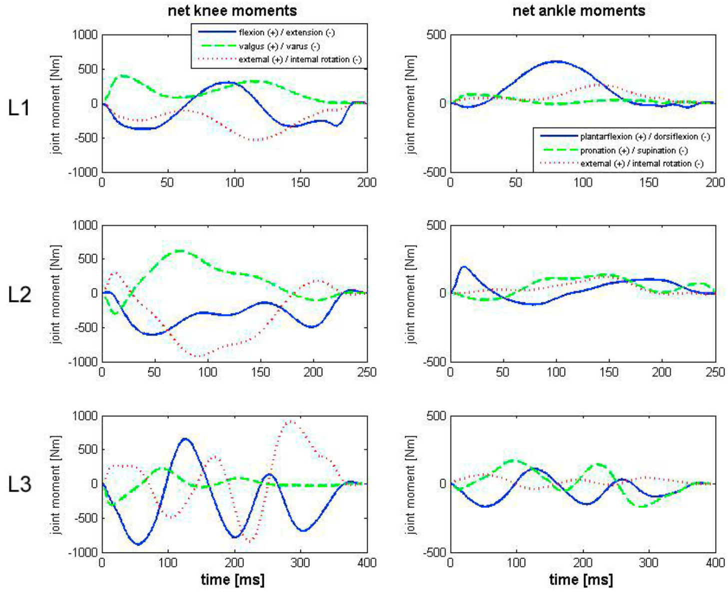

Fig. 3. Results of SIMPACK computer simulation of ACL injuries. Moments are defined from the proximal to the more distal body segment.

The comparison with simple inverse-dynamic calculation revealed principal accordance although there were several differences and deviations. The question hereby is, however, which of the models deliver the more correct results as a simple inverse dynamic approach is also prone to the limitations mentioned above. Some of the deviations can clearly be dedicated to the simulation method, e.g. the deviations in some axes after about 100 to 150ms that come from the used control law. Although, this decreases the overall quality of the results to our opinion the information value is still existent as for ACL-injury mechanisms especially the first 100 to 150 ms seems to be important.

The evaluation of the influence of the time shift observed during the validation period on the informative value of the simulation results is extremely difficult as the exact moment of injury is not known. Based on the detailed analysis of the Poser data we assume that in the situations L1 and L2 the injury occurred about 80ms to 120ms after initial ground contact. Within this period of time the deviations within the validation of the model were within an acceptable magnitude, so that these results should be reasonable. The evaluation of the injury situation L3 was much more complicated and an accurate estimation of the time of injury is almost impossible, although we believe that it could be between 50ms to 250ms after initial contact. In this case the constraints mentioned above could not be neglected. Hence, the simulation results of L3 should be treated with caution.

As almost no comparable computer simulations of 3-D movements are available checking the plausibility of the results in general is problematic. Joint moments seem to be very high, especially for the knee joint. But one has to consider that net-muscle-moments as calculated here are compositions of constraint forces and moments and muscle forces, respectively. As no subdivision into constraint loads and muscle forces have been accomplished until now, more detailed interpretation is difficult.

5. Conclusion

The aim of the study to develop and validate a new method for biomechanical computer simulation of real injury situation was obtained. The approach provides additional information compared to simple visual video inspection or Poser analysis alone. It is therefore useful for biomechanics to get a deeper understanding what happens during an ACL-injury. Of course, a number of approximations have been made and there are weaknesses which have to be improved. In conclusion the approach of combining detailed analysis of injury situations with biomechanical computer simulation is a first step towards a powerful tool for analyzing injury mechanisms. The method has the potential to improve the understanding of injury mechanisms significantly especially when the subdivision of constraint loads and muscle forces will be realized and the weaknesses named above will be eliminated.

References

[1] Chomiak J, Junge A, Peterson L, Dvorak J. Severe Injuries in Football Players. *Am J Sports Med* 2000;28(5):S58-S68.

[2] Van Mechelen W, Hlobil H, Kemper HCG. Incidence, Severity, Aetiology and Prevention of Sports Injuries – A Review of Concepts. *Sports Med* 1992;14(2):82-99.

[3] Krosshaug T, Andersen TE, Olsen O-EO, Myklebust G, Bahr R. Research approaches to describe the mechanisms of injuries in sport: limitations and possibilities. *Br J Sports Med* 2005;39:330-9.

[4] Wallrapp O, Grund T, Böhm H. Human Motion Analysis and Dynamic Simulation of Rowing. In: Goicolea JM, Cuadrado J, Orden JCG, editors. *Proceedings of MULTIBODY DYNAMICS, ECCOMAS Thematic Conference*, 2005.

[5] Gruber K, Ruder H, Denoth J, Schneider K. comparative study of impact dynamics: wobbling mass model versus rigid body models, *J Biomech* 1998;31:439-44.

[6] McLean SG, Su A, van den Bogert AJ. Development and Validation of a 3-D Model to Predict Knee Joint Loading During Dynamic Movement. *J Biomech Eng* 2003;125:864-874.

[7] Barone M, Senner V, Schaff P. ACL injury mechanism in alpine skiing: analysis of an accidental ACL rupture. In: Johnson RJ editor, *Skiing trauma and safety: 12th edition*, 1999, ASTM STP 1345, West Conshohocken, PA: American Society for Testing and Materials, p 63–81.

[8] Krosshaug T, Bahr R. A model-based image-matching technique for three-dimensional reconstruction of human motion from uncalibrated video sequences. *J Biomech* 2005;38:919-29.

[9] Hanavan E. A mathematical model of the human body. *Technical Report NO. AMRL-TR-64-102*, 1964.

[10] Zatsiorsky VM, Seluyanov VN. The mass and inertia characteristics of the main segments of the human body. In: Matsui H, Kobayashi K, editor, *Biomechanics VIII-B*, 1983, Champaign, IL, Human Kinetics, p 1152-9.

[11] Winter DA. *Biomechaniscs and Motor Control of Human Movement*. J. Wiley & Sons, New York, 1990.