A simple method to assist and optimize the TT prosthesis alignment process

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Abstract

A simple mobile kinematic measurement system acquires the knee-angular motion of the prosthetic and the contralateral leg. Using that system for fitting optimization of TT Prostheses as recommended by Blumentritt [1] provides direct and immediate hints towards the optimum described therein. The quantified knee-angle in this study is compared with Vicon Gait Lab data (Vicon 460, Oxford Metrics, GB). In this study the different kinematic and verbal reactions during the optimization process are compared. Although the response is individual verbally and by data, it can be shown that the optimum A-P position can be found in a very narrow bandwidth.

Introduction

Fitting TT Prostheses amongst proper component selection, adequate socket design and the adjustment of various degrees of freedom requires the adjustment of A-P position as well as plantar-flexion of the prosthetic foot. These sagittal plane adjustments are of high importance as the foot in its largest dimension – length – is influencing the knee stability by knee moment most. Blumentritt recommends to optimize the plantarflexion-angle by adjusting the horizontal distance of the vertical line on top of the prosthetic COP in a distance of 15mm anterior to the Compromise Pivot Axes by Nietert [2] by using L.A.S.A.R. Posture (Otto Bock HealthCare, Germany) which indicates this load-line when the patient is standing straight (see picture 1).

The appropriate A-P position is correlated to a smooth controlled initial knee-flexion and -extension after heel strike. Inappropriate A-P positions show limited knee- angle-trajectories or those deviating from the physiological knee-angular-trajectories in other aspects. Up to now, the knee-angle-trajectory is derived visually by the practitioner.

A mobile kinematic measurement system based on inertial sensors and online data processing calculates the segment angles from inertial sensor data. Furthermore it derives arguments to optimize the fitting instead relying on visual subjective observations. The verbal responses of the TT Prostheses wearers are still important, especially if they are already experienced performers of knee-flexion and -extension from heel strike to swing initiation. The graphical measurement-results are available directly after each step. In the fitting process they are assistive to optimize in the right direction and to detect when to optimize in smaller alignment changes.

Experiences of fitting TT prostheses by using the recommended alignment procedure are described by Scherer [3]. Using a simple mobile kinematic system in addition provides objective data for optimization and for a documented fitting result.
Material and Method

The inertial sensors are three orthogonal gyrometers and three orthogonal accelerometers clustered to one sensor array. Each shank and thigh on both legs is equipped with one array which is connected by a wire to the waist array and transmitter. So the right location of the leg sensors is predefined if the waist array is positioned first. As the array in a small housing are fixed to the body segment by Velcro, the application is simple and fast. During walking the processed data of the arrays is sent to a stationary PC via Bluetooth. In this study a video was captured to compare the visible knee angular characteristics with those of the data. Data capturing started when the subject was standing still. If the knee angular trajectory is not starting at zero, further knee-extension at heel-strike in comparison to standing is indicated. The first step is not displayed in the following diagrams.

Data of “inertial-measurement-based” optimization of five well experienced TT prostheses wearers is acquired for this study. The foot is adjusted by either shifting it horizontally against the socket with a shifting adapter or by tilting the connecting element between socket and foot solely. (The correct counter-tilt of the foot is determined by adjusting the 15mm knee offset by plantarflexion correction again). Criterion for the optimum A-P position is the approximation to the physiological knee angular trajectory for walking on level ground [4]. Prior to walking on the neutral aligned prostheses, the 15mm distance to the “Compromise Pivot Axes” of the knee is adjusted by plantar-flexion. The subjects are asked to walk in their preferred walking speed straight ahead. Depending to the tendency of initial knee flexion and the reaction on adjustment changes, the subjects are motivated to initial knee flexion. The motivation contains an explanation about the physiological knee angular characteristics over time by demonstrating this movement and by explaining advantages in detail. The collected data of several steps without changes in alignment are displayed in a user interface on PC screen.

The optimizations in this study, as well as recommended in clinical practice, starts with a shift of the foot in 10mm increments. If the knee-angle-trajectory appears to approximate the physiologically trajectory and the patient reports improvement, another 10mm shift in the same direction is done. Having a clear statement of the prostheses wearer or a visible deviation from previously better approximated physiological knee-angular-trajectory, the foot is shifted in the opposite direction. After each adjustment the horizontal distance to the “Compromise Pivot Axes” is registered or adjusted to 15mm. Feet which are equipped with elements reaching in the shin area cannot be altered in their angular orientation without shifting them. In those cases, to optimize plantarflexion in that degree of freedom alone, the adjustment is performed by applying wedges underneath the foot in the footwear. When reaching an A-P position where neither anterior-, nor posterior shift is improving the response while walking on level ground, further optimization can be performed in 5mm or 2.5mm increments in fast changes. The fast changes are only carried out with subjects of higher mobility grades and were not documented by gait-lab or the mobile kinematic system.

Results

The acquired data compared to gait lab data shows the same characteristics, but is not identical (see diagram 1). Deviation can be driven by different locations of the “sensors” on the body: The markers of the gait-lab acquisition system are very distant (hip, knee and Ankle) where the thigh marker of the inertial system is localised in the area of the middle of the thigh - on moving muscles. It also can be caused by plane of observation: In this gait-lab data, the sagittal knee-angle is shown, whereas the inertial-system rotates in transversal plane with the leg. However, the principle structures of the relevant signal both indicate reliably the relative knee angular motion after heel strike.

![Diagram 1](image)
Subject 1 (13cm stump length) reports noticeable pressure in the hollow of the knee at 10mm posterior position of the foot, Subject 2 (20cm stump length) in the same alignment about noticeable pressure at the anterior distal tibia end. The increased lever arm of the heel introduces a sagittal moment into the socket which is stabilized by a pair of forces of the stump (see diagram 2).

The responses of the subjects indicate that alignment influences the recognition of socket shape. It also indicates that a shape which is not received to be appropriate might see a better acceptance after optimisation of alignment.

Measured raw data of the different alignments of the individuals are overlaid for visual comparison. Different subjects show different response to the alignment optimization. Subject 1 slightly increases knee flexion after motivation to walk that way, but doubted that this easy walking is safe at outside walking (see diagram 3, 10mm posterior, after motivation). Shifting the foot 10mm anterior in relation to start condition prevents initial knee flexion. The alignment of the start condition is the final alignment too (see blue box) in diagram 3).

Subject 5 does not show initial knee-flexion in the start condition (see diagram 4). A shift of the foot by ~20mm (tilting the connecting structure between socket and foot) initiates knee-flexion spontaneously. The toe is sensed to be inactive in this configuration by the subject. Altering the shift to ~20mm anterior prevents initial knee-flexion and is found to be like going uphill. ~10mm posterior position of the foot is accepted by the subject. After explaining the benefit of knee-flexion and -extension, it is commented to be softer and easier walking. This adjustment is the final alignment.
Subject 2 shows a different response to different A-P positions of the foot (diagram 5): The major difference in the knee angular trajectory shows up in the extension phase after initial knee-flexion: The further the foot is positioned anterior, the less the extension is performed (although the toe leaver is increasing by anterior shift of the foot). The subject reports anterior distal pressure when the foot is positioned 20mm posterior to start condition. The final adjustment is 10mm posterior to start condition. Subject 3 and 4 do not show those distinct repetitive reactions on alignment changes, but clearly report what they can sense by different A-P positions and which position they prefer. Subject 3 only accepts a variation of 5 mm around the optimum.

Further Results
To inform the practitioner online and to motivate the prosthesis wearer to perform initial flexion voluntarily, audio-feedback generated by the mobile kinematic system is found to be appropriate. The audio-feedback indicates the “quality” of knee angular motions by different sounds. Prior to audio-feedback the response to A-P shift is tested in this study. Subject 1, even showing initial flexion, is preventing smooth knee-flexion. Audio-feedback based on initial flexion in this case is not appropriate for motivation. All other subjects showed initial knee-flexion either voluntary or after first A-P shift. So in this study audio-feedback is not effective.

Prosthetists being involved in this study also found the contralateral leg data to be relevant. Many places where prosthetic fitting is performed are equipped with a narrow walkway, which does not allow the sagittal observation in principle. To enrich the fitting by mobile kinetic data acquisition, can therefore lead immediately to improved results.

Conclusion
Sagittal visual analyses of the prosthesis wearer allow a decision to motivate for initial knee-flexion. Gait-Lab data as well as mobile kinetic data assist the practitioner to find the best A-P position for the individual patient. Larger studies have to be performed to define the criteria for A-P alignment recommendations based on mobile kinematic analyses more precisely. The present system already provides the ability to capture the knee angular trajectory for documentation. Independent to the willingness of the prosthesis wearer to continue initial knee-flexion and -extension after the fitting, the practitioner can optimize the A-P position following the alignment process at least during fitting. For the prosthesis wearer that alignment result is the precondition to exercise the physiological motion during daily walking voluntarily.

References

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