A comparison of the forces developed at the hip joints of ‘normal’ and total hip replacement subjects

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Introduction
Direct in-vivo measurement of hip joint contact forces has been performed (e.g. Bergmann et al., 1993) and the results have provided information on the forces that act on the femoral component of the hip joint prosthesis. However, to specify the design requirements for hip prostheses the full range of loading conditions that might be experienced must be characterised, including those for ‘normal’ subjects. Hip joint contact forces have been calculated in ‘normal’ subjects by a number of authors (Tsirakos et al., 1997). However, there is limited evidence of the comparison between the hip joint contact forces in ‘normal’ subjects and those in subjects with hip replacements.

For this study the hip joint forces of normal and total hip replacement subjects were calculated using the same protocol.

Methods
Kinematic and kinetic data were collected using a VICON motion analysis system and a Kistler force platform. These data were used in a computer model of the lower limb, which provided for 3D force and moment balance at the hip, knee and ankle (Stansfield, 2000). The knee and ankle joint force bearing structures are illustrated in Figures 1 and 2.

The hip joint was treated as a ball and socket joint. At the knee and ankle joints major force bearing structures were modelled. The knee axes were located on the tibial plateau (X pointing anteriorly, Z laterally, Y superiorly). The variation in knee centre of contact during movement was described after Nisell (1985). Supplementary forces were introduced at the knee to model other soft tissue structure contributions to joint equilibrium. At the ankle the contact between the tibia, fibula and talus was effectively modelled as a cylindrical joint with load distributed over a region ±q from the joint centre. The joint axis was defined after Isman & Inman (1968). The ankle axes were aligned with X pointing anteriorly, Z laterally and Y superiorly for the right foot.

47 muscle elements (Brand et al., 1982) were used in the model and wrapping procedures were used to ensure that muscles did not pass through underlying structures.
Muscle redundancy was accommodated by the use of a linear optimisation technique minimising the maximum muscle stress then minimising the sum of forces in the muscles and joints. Hip joint forces were calculated from the model in a femoral co-ordinate system. 5 male ‘normal’ and 5 male total hip replacement subjects were studied for walking and ascending/descending stair and ramp.

Results & Discussion
Subject characteristics are presented in Table 1. Figures 3 and 4 illustrate resultant hip joint contact forces during walking and stair ascent respectively.

Table 2 details the average and standard deviation of the maximum resultant hip joint contact force calculated for ‘normal’ subjects and subjects with prosthetic hip joints. ‘Normal’ subjects’ hip joint forces were on average higher for all activities. The average maximum force ranged from 4.99 to 7.12 for ‘normal’ subjects and 4.28 to 5.08 times body weight for subjects with prostheses. The relative magnitude of the resultant force during the different activities can be compared in Figure 5. Subjects with hip replacements exhibited lower average speed, stride length and cadence than the normal subjects (Table 3).

<table>
<thead>
<tr>
<th></th>
<th>Male normal subjects</th>
<th>Male subjects with hip replacements</th>
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</thead>
<tbody>
<tr>
<td>Age (years)</td>
<td>49.4 (5.0)</td>
<td>52.6 (6.6)</td>
</tr>
<tr>
<td>Height (cm)</td>
<td>176.9 (6.8)</td>
<td>170.8 (6.7)</td>
</tr>
<tr>
<td>Mass (kg)</td>
<td>78.5 (5.4)</td>
<td>77.8 (6.6)</td>
</tr>
<tr>
<td>Post op. (months)</td>
<td></td>
<td>18.6 (4.1)</td>
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Table 1  Subject characteristics. Average and standard deviation values are illustrated
Subjects with total hip replacements exhibited lower hip joint forces than ‘normal’ subjects. This difference may be explained by the slower speed of activity exhibited by the subjects with hip replacements compared to the normal subjects. Lower cadence coupled with shorter stride length would have been associated with lower accelerations. Lower accelerations would have been associated with lower inertial and ground reaction forces. The results indicate that this lead to lower internal forces at the hip joint contact.

The hip joint contact forces calculated for stair ascent and descent were lower than for walking and ramp ascent and descent. The stair negotiation was performed within a controlled environment with stride length fixed. This fact appears to have prevented the subjects developing high hip joint contact forces during stair ascent.

The observation that normal subjects develop higher forces at their hip joints than subjects with hip replacements suggests that using forces obtained from hip replacement subjects to define prosthesis design requirements could compromise the long term performance of the prosthesis for subjects returning to ‘normal’ levels of activity.

References

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