Supplementary article data

Development and validation of an automated and marker-free CT-based spatial analysis method (CTSA) for assessment of femoral hip implant migration

In vitro accuracy and precision comparable to that of radiostereometric analysis (RSA)

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Details of the automated analysis of CT scan images

Preprocessing and segmentation

Both baseline and follow-up CT scan images were manually cropped to exclude the pelvis, with the aim of reducing the influence of differential rotation that may occur between femur and pelvis in subsequent CT scans. Next, we automatically segmented both the femoral bone and the stem. To extract the prosthesis, a threshold was automatically determined by Otsu's method, which aims to minimize intraclass variance (Otsu 1979). That threshold was applied to the image and was followed by a morphological closing with a binary ball element with 2.0 mm radius, to obtain the complete segmentation of the prosthesis.

The prosthesis segmentation was subtracted from the original image, and the resulting image was thresholded again with Otsu's method to extract the bone. That image was morphologically eroded using a binary ball element with 0.5 mm radius and dilated using a radius of 2.0 mm to obtain the segmentation of the bone. Both segmentation masks were applied to the original images, thereby retaining the voxel intensities within the segmented objects (bone and stem), and setting the background to −1,000 HU.

Registration

After segmentation, we spatially aligned the images of both CT scan data sets. This process is called rigid registration. The baseline prosthesis image was registered to the follow-up prosthesis image using a rigid registration scheme in which the transformation was represented by:

$$T(x) = R(x–C) + D + C$$

In this equation, T is the transformation, R is the rotation matrix, C is the center of rotation, and D is the translation vector. The transformation holds 6 degrees of freedom: 3 in the translational part and 3 in the rotational part (typically the Euler angles). Similarity was measured using the mutual information metric, which was shown to lead to better registration results compared to calculating the mean squared differences (Vandemeulebroucke et al. 2013). Due to the high contrast between background and prosthesis, we used 128 histogram bins. The transform parameters were optimized using a multi-resolution approach with 6 resolution levels and an adaptive stochastic gradient descent optimizer (Klein et al. 2009). For each resolution level, 2,000 iterations were performed using 2,048 randomly chosen spatial samples.

The prosthesis-to-prosthesis registration was followed by a bone-to-bone registration performed in the same manner. To reduce computation time, bone registration was performed starting from the transformation found in the prosthesis registration and only required the use of 2 multi-resolution levels. We have previously shown that this does not introduce any bias in the results (Polfliet et al. 2015). Based on both transformation matrices obtained from the registration procedures (T_b for the bone and T_p for the prosthesis), the physical migration of the prosthesis with respect to the bone (T_m) can be calculated as (Vandemeulebroucke et al. 2013, Polfliet et al. 2015):

$$T_m = T_b^{-1} \circ T_p$$

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The choice of a coordinate system

To facilitate interpretation of the migration matrix, the data must be presented in a standardized coordinate system. In a review on RSA, Valstar et al. (2005) suggested the use of a coordinate system based on the position of the calibration box, i.e. more or less aligned with the femur. As we do not have a calibration box in our setup, we expressed the migration with respect to a coordinate system that is fixed to the implant itself. We chose the implant and not the bone, as this is the most stable, symmetrical, and easy to identify of both elements. To approximate the conventions of RSA techniques (Valstar et al. 2005), we chose the Y-axis along the longitudinal axis of the stem, while the X-axis was chosen along the projection of the principal axis of the proximal part of the prosthesis on a plane perpendicular to the Y-axis. These principal axes were automatically identified by computing the distance transform of the segmentation mask of the prosthesis, in which the most central parts of the prosthesis will receive the highest values. The highest values were identified in each slice, and a line through the 8 most distal centimeters of the stem was defined as the Y-axis. The best-fitting line for the remaining part of the segmentation (containing the proximal part of the prosthesis) was projected onto a plane perpendicular to the Y-axis and determined the orientation of the X-axis. The Z-axis was perpendicular to the X- and Y-axes, following a right-handed coordinate system placed with the origin in the tip of the stem, i.e. the intersection of the Y-axis with the most distal part of the stem.

The transformation T_{m0} that represents the migration of the prosthesis with respect to the femoral bone, expressed in the standardized coordinate system, can be written as (Polfliet et al. 2015):

$$T_{m0} = T_0^{-1} \circ T_b^{-1} \circ T_p \circ T_0$$

where T_b and T_p represent the result of the bone and prosthesis registration, respectively. T_0 represents the transformation that aligns the baseline prosthesis image with the standardized coordinate system. It is important to note that, the calculation of the relative migration of the prosthesis and the bone is independent of the coordinate system chosen. As such, it is possible to reorientate the standardized coordinate system manually to any other position.

Implementation details

The complete image analysis procedure was implemented in C++ using 2 major software packages. The Insight Segmentation and Registration Toolkit (http://www.itk.org) (Yoo et al. 2002, Ibanez et al. 2005) was used to carry out the segmentation steps and the reorientation of the coordinate system. The registration was performed using "elastix" (Klein et al. 2010).

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