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Quantitative *in vivo* analysis of the kinematics of carpal bones from three-dimensional CT images using a deformable surface model and a three-dimensional matching technique

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The purpose of this study was to obtain quantitative information of the relative displacements and rotations of the carpal bones during movement of the wrist. Axial helical CT scans were made of the wrists of 11 volunteers. The wrists were imaged in the neutral position with a conventional CT technique, and in 15–20 other postures (flexion–extension, radial–ulnar deviation) with a low-dose technique. A segmentation of the carpal bones was obtained by applying a deformable surface model to the regular-dose scan. Next, each carpal bone, the radius, and ulna in this scan was registered with the corresponding bone in each low-dose scan using a three-dimensional matching technique. A detailed definition of the surfaces of the carpal bones was obtained from the regular-dose scans. The low-dose scans provided sufficient information to obtain an accurate match of each carpal bone with its counterpart in the regular-dose scan. Accurate estimates of the relative positions and orientations of the carpal bones during flexion and deviation were obtained. This quantification will be especially useful when monitoring changes in kinematics before and after operative interventions, like mini-arthrodeses. This technique can also be applied in the quantification of the movement of other bones in the body (e.g., ankle and cortical spine). © 2000 American Association of Physicists in Medicine. [S0094-2405(00)02409-3]

Key words: carpal bones, computer animation, image registration, image segmentation, joint kinematics

I. INTRODUCTION

The wrist is a complex anatomical region comprising eight carpal bones with numerous articular surfaces, ligaments, tendons, and neurovascular structures. In common radiodiagnostic practice, wrist disorders are quantitatively analyzed by the determination of carpal bone angles from radiographic lateral projections.^{1–4} Due to the two-dimensional (2D) nature of the measurements, however, the angles show large variability and the diagnostic use is limited.

Information about the movement of the carpal bones in three dimensions was obtained in the past by tracking markers attached to the bones using x-ray stereo photogrammetry, optoelectronic stereo cinephotogrammetry, low frequency magnetic field technology, or video data, and reconstructing the geometry of the bones from sections obtained by CT, histological microsectioning, or digitization of articulation surfaces.^{5–10} Though the techniques used in these studies are very accurate, they are limited to *in vitro* applications. In

more recent *in vivo* studies, three-dimension(al) (3D) information about the relative position and orientation of the carpal bones was analyzed using an anatomical landmark¹¹ and surface registration.^{12,13} To this end, the wrist was imaged in different postures by 3D CT. The diagnostic application of these techniques is promising since the movement of the carpal bones can be quantified in 3D. However, to keep the exposure of x rays within acceptable limits, the number of CT acquisitions was limited to typically four to six different postures.

In order to obtain accurate kinematic data of different wrist motions in the same subject and to provide a realistic semicontinuous animation of these motions, we imaged the wrist in a large number of postures (typically 15–20). At each posture, which represents an increment during motion, flexion to extension for instance, the wrist was imaged by spiral CT. In the neutral position a conventional CT technique was used, and in the other postures a low-dose technique to reduce the amount of x-ray exposure. Two move-

ments of the hand were studied: radial–ulnar deviation and flexion–extension.

Previously, we developed a deformable surface model (DSM) for 3D segmentation¹⁴ of the CT scans. Using this model, accurate segmentations of the carpal bones and of the facets of the radius and ulna were obtained from the regular-dose CT scan. Next, we applied a fully automatic 3D registration algorithm based on chamfer matching¹⁵ and gray value matching in order to register each bone in the regular dose scan with its counterpart in the low-dose scans. Once the bones were registered, the translation and rotation parameters were determined from the transformation data.

To visualize the motion of the carpal bones, we have developed a visual animation tool in which each frame corresponds to a posture of the hand. The animated bones can be inspected by the observer from different viewpoints. Moreover, the user can choose one of the bones as a reference coordinate system in order to study individual carpal bone motion. In this way, subtle displacements and attitude changes between bones can be noticed. In order to make a quantitative comparison between the results of different wrists, we calculated an anatomical reference coordinate system for each wrist with axes that correspond to the natural radial–ulnar deviation, flexion–extension, and supination–pronation axes. The kinematic data were expressed in this reference system for clinical interpretation and diagnosis. The registration technique may also be used to monitor joint kinematics in patients. This is especially useful for the selection of proper therapies or to improve surgical procedures.

II. MATERIALS

A. Image acquisition

Axial helical CT scans were made of 11 volunteers and of a cadaveric specimen using spiral CT (Elcint CT-Twin/Flash, Elscint Inc., Hackensack, NJ) with a double detector array. All wrists were imaged in the neutral position with a conventional CT technique using a regular dose (R), and in 15–20 other postures (flexion–extension, radial–ulnar abduction) with a low-dose technique (L). In the cadaveric study, a medium dose (M) was also used. The imaging protocol was as follows: collimation: 2×0.5 mm, scan time: 1 s (360°), table feed 0.7 mm/s (R and M) or 2.0 mm/s (L), 120 kV, 135 mAs (R), 33 mAs (M) or 13 mAs (L). The ultra high resolution mode of the CT-Twin was used (full width at half maximum of the point spread function = 0.6 mm).¹⁶ Reconstruction matrices of 512×512 and of 340×340 were used, and reconstruction intervals of 0.3 or 0.6 mm. The reconstruction parameters are listed in Table I.

A graphical workstation (Sun SPARC 20 at 66 MHz) was used for image processing and visualization. We developed software in C and C++ to implement segmentation and registration algorithms. For 2D display and interfacing we used an image processing package (SCILIMAGE, TNO, Delft, The Netherlands). Additionally, a visualization package (AVS, Advanced Visual System Inc., Waltham, MA) was used for 3D surface rendering and animation.

TABLE I. Reconstruction parameters for regular-dose (R), middle-dose (M), and low-dose (L) CT scans. Note that the voxel size is expressed as pixel size $\times$ reconstruction interval.

protocol	(mAs)	Matrix	Voxel size (mm ³)
R	135	512×512	$0.20 \times 0.20 \times 0.3$
M	33	512×512	$0.20 \times 0.20 \times 0.3$
L ₁	13	512×512	$0.20 \times 0.20 \times 0.3$
L ₂	13	512×512	$0.20 \times 0.20 \times 0.6$
L ₃	13	340×340	$0.30 \times 0.30 \times 0.3$
L ₄	13	340×340	$0.30 \times 0.30 \times 0.6$

B. Wrist posture device

In order to position the wrist at a specific angle during radial–ulnar deviation or flexion–extension, a special posture device was designed at the Mechanical Development Department of our hospital (Academic Medical Center, Amsterdam). Using this device the angulation of the hand was established accurately and reproducibly, with increments of 5° . The apparatus consists of four parts [Figs. 1(I)–1(IV)]. The hand and forearm were placed in the center of the gantry. Since the width of the table of the CT scanner was relatively small, a wide board (I) was placed on top of the table on which the volunteer was laying. To prevent movement, the forearm was placed in a holder (II) at the top of the table board. To this board a circular arch (III) was attached in order to angulate the wrist in the lateral plane. A second arch (IV) was attached to the palmar side of the hand in order to angulate the wrist in the coronal plane. The attachment was such that the center of the wrist and that of the arches coincided. A combination of the two orthogonal angulations was established by linking the two arches at specific angles. To prevent substantial attenuation of the x-ray beam, the arches were made of Lucite.

III. REGISTRATION METHOD

The 3D registration technique consists of the following steps. First, a segmentation was obtained for each bone in the

FIG. 1. Wrist posture device. (I) Table board. (II) Holder for the forearm. (III) Flexion–extension arch connected to the table board. (IV) Radial–ulnar deviation arch attached to the palmar side of the hand.

regular-dose CT scan of the wrist in the neutral position by application of a 3D deformable triangulated surface model.¹⁴ Second, the bones in the regular-dose CT scan were registered with the corresponding bones in the low-dose CT scans of the wrist in other postures by matching the internal structure of the bones. A cost function was defined which quantifies the quality of the registration. For each bone the cost function was minimized with an iterative minimization procedure by adjusting translation and rotation parameters. Since an iterative minimization procedure was used, an initial transformation $T^{(\text{init})}$ had to be provided that transforms a bone to within the capture region of the minimization procedure. In order to obtain a relatively large region of capture chamfer matching¹⁵ was used. In a second optimization stage, gray value matching was used in an attempt to improve the match by taking the entire volumetric information of a bone into account.

In order to enable a quantitative comparison of different wrists, an anatomical reference coordinate system, derived from the distal radius in the neutral position, was chosen. This allowed us to express the translation and rotation parameters in this reference system. Despite the measures taken to fix the forearm, sometimes slight attitude changes of the radius (translation <1 cm, rotation <5°) occurred between successive scans. Therefore we aligned the radius of each scan with the radius in the neutral position by matching the radii and transforming the data to the reference system. Finally, the movement of the carpal bones could be inspected visually with an animation tool. To allow a quantitative analysis, the transformation data were expressed in the cardinal and in the finite helical axes (FHAs) representation, which is common practice in the field of biomechanics.^{17–19}

A. Chamfer matching

A bone in the image $I(x,y,z)$ of the regular scan was registered with the image $I'(x,y,z)$ of one of the low-dose scans by estimating the components of a 4×4 homogeneous transformation matrix T which minimizes a cost function $C(T)$. In homogeneous coordinates the transformation matrix T is expressed by

$$T = \begin{pmatrix} R & \vec{t} \\ 0 & 1 \end{pmatrix}, \quad (1)$$

where R is a 3×3 rotation matrix and $\vec{t}$ is a 3×1 translation vector. After minimization the features at positions $\{\vec{x}\}$ within the bone in $I(x,y,z)$ match at positions $\{\vec{x}'\}$ in $I'(x,y,z)$ with $\vec{x}' = T\vec{x}$ and $\vec{x} \equiv (\vec{x}, 1)$.

The interior of the bones exhibits an inhomogeneous texture which was very suitable to use for matching. We used the edges of the texture in the matching procedure. For that purpose edge images E and E' were obtained by thresholding images I and I' at 600 Hounsfield units and by selecting the voxels that had at least one background voxel as a neighbor. For edge image E the positions $\{\vec{x}\}$ of the “edge voxels” inside each bone were used in the calculation of the cost function. The other edge image E' served as a template for which a distance transformation D' was calculated (Fig. 2)

FIG. 2. Distance transformation. A coronal image slice of a low-dose CT scan is shown (left), from which a 3D edge image is derived (middle). From the edge image a 3D distance transformation is calculated (right). Dark areas indicate pixels which are located near the edges, while the distance to the nearest edge is high in bright areas.

by applying the city-block distance transformation.¹⁵ The cost function C_{dist} was defined as the average of the distance transformation at coordinates $\{\vec{x}'\}$, i.e., the average distance of the transformed edge voxels of E to the edges in E' :

$$C_{\text{dist}} = \langle D'(\vec{x}') \rangle. \quad (2)$$

The values $D'(\vec{x}')$ were obtained by trilinear interpolation. In order to deal with the situation that not exactly the same part of the bone was contained in I and I' , which is the case for the radius and ulna, the points of $\{\vec{x}'\}$ that were outside I' during optimization were not taken into consideration in that optimization stage.

To find the minimum value of the cost function we applied the *downhill simplex method* due to Nelder and Mead.²⁰ The code of this algorithm was obtained from the *Numerical Recipes in C*.²¹

Initialization. The transformation matrix T contains six parameters: three translations (t_x, t_y, t_z) in the X , Y , and Z directions and three rotations ($\theta_x, \theta_y, \theta_z$) around the X , Y , and Z axes. The simplex algorithm requires an initial estimate of the transformation matrix $T^{(\text{init})}$, and an estimate of the order of magnitude of the deviation to be expected from the initial values, for which we chose $\Delta t = 2$ mm and $\Delta \theta = 5^\circ$. A fully automatic procedure was obtained as follows. Let the 3D image sequence corresponding to a set of positions of the wrist, e.g., from flexion to extension, be denoted by I_j , where the index j indicates the position of the wrist ($j = \dots, -3, -2, -1, 0, 1, 2, 3, \dots$), with $j = 0$ for the neutral position. $T_{C_i,j}$, $T_{R,j}$, $T_{U,j}$, and $T_{\Sigma,j}$ indicate transformation matrices for the carpal bone i (C_i), the radius (R), the ulna (U), or the carpus as a whole (Σ) containing the eight carpal bones, which register the relevant element (C_i , R , U , or Σ) in image I_j with its counterpart in image I_0 . First the radius was registered. For registration with $j = \pm 1$, as the initialization matrix the unit matrix I was taken; otherwise we initialized with the transformation matrix of the nearest position. Next the carpus was registered. For $j = \pm 1$ we initialized with the transformation matrix of the radius in the same image; otherwise the realigned transformation matrix of the nearest carpus was taken. The carpal bones and the ulna were initialized with the transformation matrices of the corresponding carpus and radius, respectively. Summarizing, the rules for initialization were

$$\begin{aligned}
T_{R,j}^{(\text{init})} &= \begin{cases} \mathbb{I}, & j = \pm 1 \\ T_{R,j \pm 1}, & j \in \mathbb{Z}^+ / \{\mp 1\}, \end{cases} \\
T_{\Sigma,j}^{(\text{init})} &= \begin{cases} T_{R,j}, & j = \pm 1 \\ T_{\Sigma,j \pm 1} T_{R,j \pm 1}^{-1} T_{R,j}, & j \in \mathbb{Z}^+ / \{\mp 1\}, \end{cases} \\
T_{U,j}^{(\text{init})} &= T_{R,j}, j \in \mathbb{Z}^+, \\
T_{C,j}^{(\text{init})} &= T_{\Sigma,j}, j \in \mathbb{Z}^+.
\end{aligned} \quad (3)$$

In only isolated cases this scheme produced an initialization that failed to converge to a correct solution. The initialization was performed manually in those cases.

B. Gray value matching

In an attempt to improve the quality of the match we introduced a second optimization stage in which we used two other cost functions. These cost functions were based on a comparison of the image intensities $I(\vec{x})$ and $I'(\vec{x}')$ by taking either the rms difference (C_{rms}) or by calculating the correlation coefficient (C_{corr}):

$$\begin{aligned}
C_{\text{rms}} &= \sqrt{\langle (I(\vec{x}) - I'(\vec{x}'))^2 \rangle}, \\
C_{\text{corr}} &= \frac{-\langle (I(\vec{x}) - \langle I(\vec{x}) \rangle)(I'(\vec{x}') - \langle I'(\vec{x}') \rangle) \rangle}{\sqrt{\langle (I(\vec{x}) - \langle I(\vec{x}) \rangle)^2 \rangle \langle (I'(\vec{x}') - \langle I'(\vec{x}') \rangle)^2 \rangle}}.
\end{aligned} \quad (4)$$

In the calculation of these cost functions in principle the voxels $\{\vec{x}\}$ with a value greater than 600 Hounsfield units of image I were used. To reduce the computational burden, however, 10% of these points was chosen at random with a maximum of 50 000 points. The values $I'(\vec{x}')$ were obtained by trilinear interpolation. The transformation obtained with chamfer matching was used as initialization for this stage and we took $\Delta t = 1$ mm and $\Delta \theta = 1^\circ$.

C. Verification methods

As a measure of the quality of the match, the correlation between the image intensity I_0 of all voxels within each bone and the image intensity $I_{j \neq 0}$ at the corresponding transformed positions was calculated. When the correlation was less than 0.7 we inspected the registration visually by comparison of three orthogonal slices through the bones in image I_0 and the corresponding matched slices in the images $I_{j \neq 0}$. The corresponding slices were mixed in a chessboardlike way¹⁵ (an example is shown in Sec. IV) and viewed in a movie loop.

D. Coordinate transformations

After the transformations were estimated and their correctness was verified, two coordinate transformations were applied. First, each scan was aligned with the scan in the neutral position using the radius in both scans as reference:

$$\begin{aligned}
T'_{R,j} &= \mathbb{I}, \\
T'_{B,j} &= T_{R,j}^{-1} T_{B,j},
\end{aligned} \quad (5)$$

where B denotes the carpal bones and the ulna.

FIG. 3. Anatomic reference coordinate system showing the flexion-extension (X), supi-pronation (Y), and radial-ulnar deviation (Z) axes. This system is determined from the surface of the radius, which was obtained from a match of two sets of scans: a low-dose scan of the forearm (transparent), a regular-dose scan of the wrist. The surfaces of the carpal bones, radius and ulna of the regular-dose CT scan are also shown (right wrist, dorsal side on top).

An anatomic reference coordinate system was required in order to represent the transformations in terms of translations along and rotations about the flexion-extension, the supi-pronation, and the radial-ulnar deviation axes, denoted by the X , Y , and Z axes, respectively (Fig. 3). In practice, it turned out that the distal part of the radius (≈ 5 cm) could be used to determine a reference system. Since only a small portion (< 2 cm) of the radius and ulna was imaged in the multiposture CT scans, an additional (low-dose) CT scan was taken of the forearm (Fig. 4 bottom). The radius and ulna of the low-dose scan were matched and aligned with the top of the radius and ulna in the scan of the wrist in the neutral position. The principal axes through the radius in the scan of the forearm were used in the determination of the coordinate system. They correspond to the (orthonormal) eigenvectors of a 3×3 symmetric matrix $\mathbf{P}^T \mathbf{P}$ derived from a matrix $\mathbf{P}$ which holds the positions of the vertices of the segmented surface of the radius. The eigenvectors were calculated by a *Jacobi* method.²²

Next, the second coordinate transformation was performed. First, the longitudinal Y axis ($\hat{e}_y$) was determined from the eigenvector with the highest eigenvalue; it was directed distally along the radius. Second, the X axis ($\hat{e}_x$) was determined from the remaining eigenvectors; this vector was roughly parallel to the vector connecting the centers of the radius and ulna, and directed toward the ulnar center. The Z axis ($\hat{e}_z$) followed from the outer product of the former axes. The origin of the system was chosen to coincide with the center $\vec{c}_R$ of the (aligned) distal radial surface extracted from

FIG. 4. Coronal cross sections in 3D axial helical CT scans of the wrist: a regular-dose scan (protocol R) of the neutral position (top), a low-dose scan (protocol L₄) of another posture (middle), and of a low-dose scan (protocol L₁) of the lower part of the forearm (bottom).

the scan taken from the forearm. Summarizing, the (inverse) system transformation matrix is given by

$$U_{\text{sys}}^{-1} = \begin{pmatrix} \hat{e}_x & \hat{e}_y & \hat{e}_z & \vec{c}_R \\ 0 & & & 1 \end{pmatrix}. \quad (6)$$

Finally, all bone transformations were expressed in this system:

$$T''_{B,j} = U_{\text{sys}}^{-1} T'_{B,j} U_{\text{sys}}, \quad (7)$$

where B denotes the carpal bones, the radius, and the ulna.

E. Finite helical axis representation

We expressed the transformations $\{T''\}$ in finite helical axis (FHA) parameters. This representation is especially convenient to express the rotation parameters in since the rota-

FIG. 5. Finite helical axis parameters to express transformation. One of the carpal bones is shown at two different positions ($\vec{p}_1$ and $\vec{p}_2$) with two different orientations. The helical axis is the line through a point $\vec{s}$ which points into the direction $\hat{n}$. The transformation between two bones can be obtained by shifting the bone over a distance t_h along the helical axis and rotating the bone by an angle θ about the helical axis. Courtesy of H.J. Woltring *et al.*

tion is characterized by only one axial rotation angle and the direction of the screw axis, instead of three (arbitrary) Euler angles in a cardinal representation.^{18,23}

The helical transformation is expressed in terms of a rotation about a helical axis by an angle θ , and a translation or shift along this axis over a distance t_h (Fig. 5). The helical axis is characterized in terms of a unit vector $\hat{n}$ and by the position $\vec{s}$ of some point on this axis. Following Woltring *et al.*¹⁸ and Spoor and Veldpaus,¹⁷ $\hat{n}$ and θ are derived from the following relations:

$$\cos \theta \mathbf{I} + (1 - \cos \theta) \hat{n} \hat{n}^T = \frac{1}{2} (R + R^T), \quad (8)$$

$$\sin \theta A\{\hat{n}\} = \frac{1}{2} (R - R^T),$$

with $A\vec{v} = \hat{n} \times \vec{v}$ and $0 < \theta < \pi$, where $A = A\{\hat{n}\}$ represents a skew-symmetric matrix in terms of its axial vector $\hat{n}$. After Woltring *et al.*¹⁸ we chose $\vec{s}$ to be the projection of the midpoint $\vec{p}$ on a finite translation vector $\vec{d}$ from position $\vec{p}_1$ to $\vec{p}_2$ of a bone before and after transformation, from which followed

$$\begin{aligned} \vec{p} &\equiv \frac{1}{2} (\vec{c} + \vec{c}'), \\ \vec{d} &\equiv \vec{c}' - \vec{c} = \vec{t}, \end{aligned} \quad (9)$$

$$\vec{s} = \vec{p} + \{2 \tan(\frac{1}{2} \theta)\}^{-1} \hat{n} \times \vec{d},$$

$$t = \hat{n}^T \vec{d},$$

FIG. 6. The influence of the x-ray exposure and reconstruction parameters on the accuracy of the match (R–R, R–M, R–L₁–R–L₄; see Table I) for the three cost functions C_{dist} (1), C_{rms} (2), and C_{corr} (3). Averages for errors in rotation ($\Delta\theta$) and translation (Δt), and the average correlation (c) between image intensities of the regular-dose scan vs medium- and low-dose scans are shown.

where we took $\vec{p}_1$ and $\vec{p}_2$ to be the bone centers $\vec{c}$ and $\vec{c}'$ before and after transformation, respectively.

IV. RESULTS

A. Validation of the method

To quantify the influence of the x-ray exposure on the accuracy of the measurements, we applied the matching technique to 3D CT data that were obtained from a cadaveric wrist in different positions. We also determined the influence of the matrix size and reconstruction interval used to reconstruct the CT data.

Our validation consisted of the determination of the variation in the translation and rotation parameters between the bones of the wrist when all bones were translated and rotated in an identical way. To this end a cadaveric wrist was placed in three different positions, indicated by (I), (II), and (III). Care was taken to move the hand as a whole, and as a consequence all bones underwent the same transformation. Of each the position CT scans were obtained using three different techniques: 135 mAs (regular dose, R), 33 mAs (medium dose, M) and 13 mAs (low dose, L). All scans were reconstructed with a 512×512 matrix; for the low-dose scans also a 340×340 matrix and a larger reconstruction interval were used (Table I). As a result, six 3D CT images were obtained for each position.

From each regular-dose CT scan a segmentation was obtained of the eight carpal bones. Next, each carpal bone in one of the regular-dose scans was registered with its counterpart in all other scans using chamfer matching and gray value matching with two cost functions. For each bone, 153 registrations (with transformation matrices $\{T_{\text{bone}}\}$) were performed: 17 (2 regular-dose scans in the other two positions $+5 \times 3$ medium- and low-dose scans) $\times 3$ (matchings) $\times 3$ (the number of regular-dose scans).

For practical reasons we were unable to determine the transformations that transform the wrist from position i to j ($i, j = \text{I, II, III}$) by measuring the translations and rotations by external means. Instead we approximated this transformation with the transformation of the whole carpus (Σ) from the regular-dose scan $R^{(i)}$ to regular-dose scan $R^{(j)}$, indicated by T_{Σ} . This approximation was calculated using the cost function that yielded the highest correlation c in image intensity,

taking into account the image intensity of all voxels within each segmented bone of the carpus of the regular-dose scan and the image intensity at the corresponding transformed positions using trilinear interpolation. The six matching procedures for registration of the regular-dose carpi were initialized by taking the identity transformation matrix $\mathbf{I}$. The matching of the individual bones was initialized with the transformation matrices of the respective carpi.

The following error analysis was performed on the transformations $\{T_{\text{bone}}\}$. For each carpal bone we defined an error matrix T_{err} as follows:

$$T_{\text{err}} = T_{\Sigma}^{-1} T_{\text{bone}}. \quad (10)$$

For an exact match, $T_{\text{err}} = \mathbf{I}$; otherwise, T_{err} had a slight deviation from $\mathbf{I}$ due to a translation error Δt and (axial) rotation error $\Delta\theta$. Also, the correlation c was calculated for each bone individually. In Fig. 6 the average values of $\Delta\theta$, Δt , and c for the R–R, R–M, and R–L₁–R–L₄ registrations are presented. As expected, the errors slightly increased when the x-ray exposure was reduced. The correlation dropped from 0.95 for R–R to 0.90 for R–L registrations. With respect to the different techniques used for the low-dose scans, the accuracy of the rotation angle θ was slightly less when a 340×340 matrix was used, while the accuracy of the translation turned out to be independent of the matrix size. The increment between the slices did not degrade the accuracy of the match in low-dose scans. For all registrations the mean error in rotation was less than 0.4° and in translation less than 0.5 mm. The cost function C_{rms} proved to give the best overall performance. Consequently, this cost function was used in the second optimization stage in the rest of the study.

B. Registration of the carpal bones, the radius, and ulna in 3D CT images of the wrist

We applied the matching technique to the right wrist of eleven of volunteers. Each wrist was imaged in 13–17 postures by spiral CT during radial–ulnar deviation motion with angular increments of 5° . From seven wrists helical CT scans were also obtained in 5–6 postures during flexion–extension motion with angular increments of 15° . In order to keep the total examination time within acceptable limits we used imaging protocol L₄. In addition a low-dose scan (protocol L₁) was taken of the lower part of the forearm. An example of a

FIG. 7. Registration of the scaphoid (left) and the hamate (right) of one volunteer. Coronal cross sections (A) through the bones in the regular-dose scan taken from the neutral position of the wrist and the matched counterparts (B) in 30° radial deviation are shown. In order to facilitate the judgement of the quality of the match, the original and matched cross sections are displayed intermingled in a chessboardlike fashion (C).

regular-dose scan taken in the neutral position, and of a low-dose scan taken of another posture and of the forearm, is shown in Fig. 4.

After data acquisition a surface segmentation was obtained of the wrist in the neutral position by application of a DSM¹⁴ to each carpal bone and the distal facets of the radius and ulna. The radius and ulna were also segmented in the low-dose scan taken from the forearm. Next, the carpal

bones, radius, and ulna in the regular-dose scan were registered with those in the low-dose scans. The radius was also registered with the low-dose scan of the forearm to obtain a more accurate anatomical reference system by the principal axes method as explained in Sec. III D. An example of the segmentation and the reference system derived from the radius is shown in Fig. 3.

We checked the quality of the match for each bone by the verification methods as explained in Sec. III C. In three cases the registration was incorrect due to a mismatch of the radius caused by a different starting position of the scan table for two successive scans. This caused the radius to be displaced out of the region of capture (≈ 1 cm) of the minimization procedure. In these cases manual initializations provided correct registrations.

As a final step we applied the coordinate transformations as explained in Sec. III D and the transformations of the individual bones were expressed in basic anatomical directions. The quality of the match of two bones is shown in Fig. 7 by using chessboard alternations. An example of the FHA parameters describing the movement of a capitate during radial–ulnar deviation and flexion–extension is given in Table II. If the capitate would follow the induced rotations perfectly, one expects that the values of ϕ_z at the top of Table II equal the values of θ (and n_z equals 1), and the same holds for the values of ϕ_x and θ (and n_x) at the bottom of Table II. To a large extent this is indeed the case for the flexion–extension movement (Table II bottom). The movement during deviation, however, is dispersed over more directions (Table II top). The finite helical axes of two bones are shown in Fig. 8.

TABLE II. FHA parameters describing the movement of the capitate during radial–ulnar deviation (top) and flexion–extension (bottom) of the wrist of one volunteer. The value of θ represents the global wrist angulation around the Z axis (top) and around the X axis (bottom). The components of the helical axis (n_x, n_y, n_z), the helical rotation angle (ϕ), its angular projections (ϕ_x, ϕ_y, ϕ_z) on the helical axis, and the longitudinal helical translation (t_h) are listed. Angles are expressed in deg. and translation is expressed in millimeters.

θ	n_x	n_y	n_z	ϕ	ϕ_x	ϕ_y	ϕ_z	t_h
30	-0.21	0.42	0.88	22.1	-4.7	9.2	19.6	0.9
25	-0.13	0.47	0.87	20.1	-2.6	9.4	17.6	0.8
20	-0.20	0.42	0.88	18.1	-3.6	7.7	16.0	0.8
15	-0.18	0.57	0.80	14.2	-2.6	8.0	11.4	-0.4
10	-0.08	0.58	0.81	10.9	-0.9	6.3	8.8	-0.5
5	0.01	0.56	0.83	6.1	0.0	3.4	5.1	-0.2
-5	0.55	-0.43	-0.72	5.1	2.8	-2.2	-3.7	-1.7
-10	0.38	-0.39	-0.84	12.0	4.5	-4.6	-10.1	-2.9
-15	0.32	-0.36	-0.87	17.1	5.6	-6.2	-15.0	-4.1
-20	0.42	-0.32	-0.85	22.4	9.4	-7.1	-19.0	-5.6
-25	0.37	-0.33	-0.87	27.2	10.2	-8.9	-23.6	-6.5
-30	0.51	-0.36	-0.78	35.4	18.2	-12.6	-27.6	-8.4
-60	-0.94	-0.16	-0.32	67.7	-63.3	-10.7	-21.4	0.6
-45	-0.94	-0.15	-0.30	53.1	-50.0	-7.8	-16.0	0.3
-30	-0.94	-0.13	-0.30	36.4	-34.3	-4.8	-11.0	0.2
-15	-0.90	-0.19	-0.39	18.4	-16.6	-3.4	-7.1	-0.1
15	1.00	0.05	-0.00	12.0	12.0	0.6	-0.0	-0.2
30	0.98	0.10	0.17	26.9	26.4	2.8	4.7	0.2
45	0.98	0.10	0.17	42.7	41.9	4.1	7.1	0.4
60	0.98	0.10	0.19	60.2	58.7	6.0	11.7	0.3

FIG. 8. (Color) Finite helical axes display of the radial-ulnar deviation (blue) of the capitate and lunate in coronal (top) and lateral (bottom) view. During deviation the lunate moves in the coronal plane and rotates to the palmar and dorsal side of the hand on top of the radius and ulna (transparent gray), which gives rise to laterally directed helical axes (rose). The rotation of the capitate, however, is opposed to that of the lunate. As a result the helical axes of the capitate (red) are perpendicular to the coronal plane.

V. DISCUSSION

Accurate estimates of the kinematics of the carpal bones could be obtained *in vivo* by application of the matching

technique described in this paper. Since information of the entire volume of the bones was used in the matching algorithm, the translation and rotation parameters could be estimated much more accurately than algorithms based on information of the surface only, such as the previously mentioned anatomical landmark technique, or in case of surface registration. This was important since the FHA parameters are very sensitive to errors in rotation angles, especially when small increments in motion are applied.^{18,24} Another advantage of the technique described here was that no repeated segmentation or manual landmark detection was required, which dramatically reduced the amount of human operator intervention. The matching procedure was fully automatic. The initialization scheme in Eq. (3) was sufficiently robust. Only in 3 out of more than 1700 cases did the scheme fail and operator intervention was needed for initialization. The registration method can also be applied for the kinematic analysis of other joints or the cortical spine.

In this study chamfer matching was used which features a relatively large capture range,¹⁵ followed by gray value matching to improve the accuracy of the fit. Most accurate results were obtained when in this second stage the rms cost function was used (Fig. 6). In recent literature a large number of image matching techniques were described. Most of these techniques, such as image registration based on mutual information,^{25–27} concern intermodality registration, and probably do not have particular advantages when applied to images obtained with the same modality. The automated image registration method of Woods *et al.*²⁸ can be used for both intra- and intermodality registration. For intramodality registration different cost functions were used, one of which is identical to the rms cost function used in this study. In a validation study²⁸ another of these cost functions, based on the uniformity of the ratio of two images, performed marginally better. In our situation, however, the quality of the registration is mainly determined by the imaging parameters (cf. Fig. 6), and only to a small extent by the cost function used in the final optimization.

In order to reduce the amount of x-ray exposure a low-dose CT technique was used. Using this technique the wrist could be imaged in a large number of postures representing increments during motion with only a limited radiation dose. The effective dose for each scan is proportional to the mAs value that was used (Table I). The total effective dose for all CT examinations was estimated to be less than 0.1 mSv. An error analysis in which the influence of the x-ray dose, the size of the reconstruction matrix, and the reconstruction interval was determined, demonstrated that the accuracy of the registration procedure for the low-dose scans is about 0.4 mm and 0.2° for translation and rotation, respectively, using the rms cost function and full-size matrices. Figure 6 shows that the rotation error is virtually independent of the radiation dose, while a smaller translation error could have been obtained with a regular-dose CT technique.

After a few spiral CT scans, the raw data storage of our CT scanner was full and the images had to be reconstructed prior to new acquisitions. This caused a substantial increase in the examination time. When a complete study had been

FIG. 9. Wrist motion from radial (top-left) to ulnar (bottom-right) deviation. Each frame of the animation corresponds to a different posture during motion.

executed using full-size matrices and the 0.3 mm reconstruction interval, the examination time would have taken more than 4 h, which is clearly unacceptable. Therefore, in the present study we used a 340×340 instead of a 512×512 matrix, and a reconstruction interval of 0.6 instead of 0.3 mm. This resulted in a more acceptable duration of about 2 h for the complete study. Since the different pixel sizes used (0.2×0.2 and 0.3×0.3 mm²) are smaller than the resolution of the CT scanner the use of the 340×340 matrix only slightly reduced the accuracy of the match. The use of a full matrix size, however, is to be preferred. The use of a larger reconstruction interval hardly influenced the accuracy (Fig. 6). This is expected, as the axial resolution of the scans is reduced in comparison to the regular- and medium-dose

scans as a result of the larger table feed (2.0 mm/s) that was used for the low-dose scans. We note that on various state-of-the-art CT scanning systems the storage limitations are absent and the above-mentioned considerations no longer apply. In this situation the examination can be performed within 30 min and all reconstructions can be made with full-size matrices and the smaller reconstruction interval after the examination is completed.

Next to translation and rotation, scaling may be useful in situations when the dimensions of the images are not exactly known. We have tried the matching procedure using three scaling parameters as well. It turned out that a slight decrease of the minimum of the cost function was obtained, but that the scaling factors were not systematically different from

unity and that these deviations were very small. Therefore we refrained from scaling.

Magnetic resonance imaging (MRI) offers the ability to image the wrist in 3D as well and has the advantage that it is completely harmless. However, MRI has some practical drawbacks. Inside the MR scanning system there is minimal space available for the placement of a posture device. In addition, longer acquisition times are required to obtain 3D images with a signal-to-noise ratio and resolution comparable to that of CT. Moreover, in MR images distortion may be present due to magnetic field inhomogeneity with a possible unfavorable effect on the accuracy of the measurements. Also, distortions due to chemical shift effects or artifacts due to aliasing or to postoperative susceptibility changes of tissues may counteract the accuracy.²⁹

The registration can be used to study both normal and pathological wrist motion by viewing the bones using a visual animation tool that we developed especially for "carpal navigation." For this animation the transformation sequence $\{T''\}$ [see Eq. (7)] is imported by the tool as well as the segmented surfaces of the bones. At each posture the surfaces of the bones are visualized according to their calculated transformation. The user can interact with the movie by making a selection of bones to inspect or can choose to fix the coordinate system of a specific bone to study individual bone motion. Optionally, the surfaces can be represented by their principal axes in order to obtain an improved overview when bones would be obscured otherwise. Also the helical axes can be viewed in conjunction with the animated bones. Some frames of an animation of a moving wrist are displayed in Fig. 9. The animation is of vital importance in understanding the complex movements of the bones during different motions of the wrist.

From visual inspections by orthopaedic surgeons, it turned out that the principal axes through the distal part of the radius are well suited for an anatomical reference system. By the use of this reference system the transformations of the bones can be expressed along radial-ulnar deviation, flexion-extension, and supi-pronation axes using FHA or cardinal representation. With this reference system an accurate statistical analysis of individual wrists can be performed.

At present we are collecting more data on normal, healthy volunteers in order to create a database for normal carpal kinematics *in vivo* for future reference. Until now, it has been very difficult to detect post-traumatic carpal instability caused by ligament disruption at an early stage. Patients usually appear late, as no abnormalities are apparent on early x-ray images. The way to the correct analysis is often long and includes more than one diagnostic modality, such as video fluoroscopy, arthrography, MRI, and/or arthroscopy. Early detection, however, is mandatory for the most favorable therapeutic option, i.e., restoring normal anatomy. It is our ultimate goal to detect abnormal carpal kinematics at a very early stage by using the technique described in this article. In this way the *in vivo* kinematics of the injured wrist can be compared with a large database of normal *in vivo* carpal kinematics without additional diagnostic examinations. We expect that a ligament lesion which possibly un-

derlies the carpal instability will be detected with high specificity and sensitivity.

Recently, we have started to study altered carpal kinematics after operative procedures. This could provide valuable information on the long-term results of these operations and possibly predict results of new techniques in the fast evolving field of wrist surgery.

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