Wrists in space: deformable models for segmentation and matching techniques for registration of 3-D MR and CT images of the wrist

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MEDICAL IMAGING AND IMAGE ANALYSIS
Figure 1: One of the first x-ray photographs made by Wilhelm Röntgen (1895). It shows the hand of his wife Berta adorned with a ring.
1.1 The History of Medical Imaging

PHOTOGRAPHY was not invented from one day on the other. About 350 years BC it was Aristoteles who suggested the principle of the camera obscura, the creation of an image by projecting light through a pinhole onto a plane. Many people tried to record images obtained by such a camera. During ages, the only way to achieve this was by painting or drawing. This lasted until 1822, when the Frenchman Joseph Nicéphore Nièpce invented a way to record pictures. He coated tin plates with Syrian asphalt for this purpose. Four years later he was the first to make a picture with a camera. He called his technique heliography [1, 2]. From that time, the development of photographic cameras and imagery went on rapidly and the way was also open to image the exterior of the human body. In the beginning long exposure times and large reflection screens were required. Most of all portraits and ‘whole-body images’ were obtained in this way.

The German physicist Wilhelm Conrad Röntgen discovered the existence of x-rays (so named by Röntgen because of their unknown nature) in 1895. He made the first x-ray photographs, one of which shows his wife’s hand with ring (Fig. 1). It is remarkable that the first known recording of an x-ray image is that of a hand, medical imagery of the wrist being the subject of this thesis. At the same time it was also the first successful attempt to image the interior of the human body by non-invasive means. Within months of his discovery medical practitioners began using x-rays to diagnose bone fractures. It appeared that the mysterious rays – nowadays known to be electromagnetic radiation with extremely short wavelength – are attenuated by dense structure of bones leaving shadows on the film. Softer tissues are more easily penetrated and therefore do not appear distinctly in pictures.

With the discovery of x-rays, a new specialty of its own right was born in medicine: medical imaging. Since then, many other techniques have been investigated and developed to peek inside the human body.

The first radiological study of a wrist was reported by Bryce [3] in 1896 and it can be regarded as the first published account of normal wrist kinematics. The first xeroradiographic image was published by Righi in 1907 [4]. He showed the principles of creating images of charged particles on the basis of ionizing x-rays passing through a hand (Fig. 2). This electrostatic technique has the advantage that the images do not have to be developed as is the case with x-ray imaging on film. Therefore, it was especially suitable for mobile work. It has been applied, for example, for mammography to detect breast cancer. For technical reasons it lasted until the mid 60s, however, to be applied on a large scale. Nowadays xeroradiography is obsolete.
Figure 2: The first xeroradiographic image made by Augusto Righi (1903). It shows the electric shadow image (left) of his hand obtained by x-rays. The setup of his experiment (right) consists of a x-ray source (C) which is placed above a metal frame (AB). The hand is placed on an aluminium plate attached to this frame. An ebonite disk (DE) is placed on a metal plate (FG) which is connected by an isolated part (IK) to the plate (MN) of the apparatus, which is grounded to the earth. The ebonite plate is negatively charged by an electrostatic generator. When the air is ionized by x-rays, the positively charged ions are attracted to the ebonite plate. Due to the differences in transmission of x-rays through the hand, the electric charge density varies over the ebonite plate, giving rise to electric shadows. These can be made visible by sprinkling powder that consists of a mixture of lead-mimium and sulfate, which leave red and yellow shades.
The $\gamma$-imaging of the distribution of radio-pharmaceuticals in the human body was a logical extension of the counting techniques for detecting ionizing radiation, which go back to the invention of the Geiger-Müller tube in 1929. This imaging technique is used to obtain functional information of the body and to detect tumours and diseases. The first point-by-point image was constructed by Ansell and Rotblat in 1948, which might be regarded as the first clinical nuclear medicine scan [5]. The concept of the $\gamma$-camera was due to Copeland and Benjamin who used a photographic plate in a pinhole camera in 1949 [6]. It lasted until the late 70s that the first single-photon emission computed tomography (SPECT) and positron emission tomography (PET) scanners became clinically available.

Ultrasonic imaging based on the pulse-echo principle became possible due to the development of fast electronic pulse technology during the Second World War. It is the second most frequently used investigative imaging technique after x-radiology. Initially, ultrasound was widely applied to investigate the pregnant abdomen, but at present it is also used for other diagnostic applications.

At the British Institute of Radiology annual conference in 1972, Hounsfield announced a machine used to perform x-ray computed tomography (CT) in a clinical environment. This has been considered as the greatest step forward in radiology since Röntgen’s discovery, which yielded him – together with Cormack, another pioneer in this field, the Nobel Prize for Physiology and Medicine in 1979. A CT scanner reconstructs an image slice of the body from transmission measurements performed from many angles by revolving an x-ray tube and a number of detectors around the patient. Consequently, a cross-sectional image is obtained instead of a projection image, which requires additional knowledge for proper diagnostic interpretation. Advanced reconstruction techniques made it possible to show subtle differences between tissues for proper diagnostic interpretation.

The discovery of the phenomenon of nuclear magnetic resonance (NMR) was announced simultaneously and independently in 1946 by groups headed by Bloch and by Purcell, who shared a Nobel Prize. NMR came into clinical use less than a decade after Lauterbur [7] published the first NMR-images. In 1977 the first NMR-image of a hand (Fig. 3) was reported by Andrew et al. [8], who used a 0.7 T magnet. Due to its ability to depict soft tissues in high contrast, magnetic resonance imaging (MRI) has proven to be an effective means of examining abnormal human pathology, e.g. in the brain and the spinal cord. Although MRI involves large investments, it has experienced a spectacular development in the last decade and has become of major interest in medicine.
1.2 Medical Image Analysis

After this historical introduction I shall now focus on the present situation. We have arrived at a new era in medicine, in which high-tech digital medical imaging becomes available in hospitals. This transition has an enormous impact on diagnostic radiology and other medical specialisms. Different medical departments are connected by computer networks. Images which are obtained by modalities such as SPECT or CT are stored on massive image archives, and can be retrieved from and presented at a single computer workstation. Different image manipulation techniques, such as image magnification or contrast enhancement, and multi-planar image reconstruction offer the radiologist an optimal view of these images.

By grouping the image into parts that are homogeneous with respect to one or more characteristics, a segmented image can be obtained which is suited for image measurement or visualization. From the segmented image, basic quantities, such as the volume of organs and tumours or the diameter of aneurysms, can be measured.

Volumetric scans of patients can be considered as stacked sliced data that
Figure 4: Cross-sections of the wrist in the coronal plane (left) and the axial plane (right). The wrist is a complex anatomical region comprising eight carpal bones, hamate (H), capitate (C), trapezoid (Zd), trapezium (Zm), triquetrum (T), lunate (L), scaphoid (S), and pisiform, which articulate with five metacarpal bones (MC\textsubscript{1} to MC\textsubscript{5}), radius (R), and ulna (U). In the cross-sections numerous articular surfaces, ligaments, tendons, and neurovascular structures appear. Courtesy of J.M.G. Kauer and J.M.F. Landsmeer.

constitute a three-dimensional (3-D) image. The 3-D anatomical structure can be visualized by projection (volume rendering) and shading (surface rendering) techniques. It can be inspected from different viewpoints by the observer and its appearance can be altered by changing attributes like reflectance, opacity and texture, to aid the imaginative powers of the radiologist.

The role of medical imaging has expanded beyond the pure diagnosis of pathology, however. It is becoming a tool for surgical planning and simulation, intra-operative navigation, radiotherapy planning, and for monitoring the process of disease. Although modern imaging devices provide exceptional views of the internal anatomy, the use of computers to quantify and analyze the embedded structures with any accuracy is limited [9].

1.3 Purpose of the Study

The wrist is a complex anatomical region comprising eight carpal bones with numerous articular surfaces, ligaments, tendons, and neurovascular structures (Fig. 4). Wrist pain, with or without dysfunction, is a common complaint, but a correct diagnosis is difficult as the clinical signs are often subtle. The way to a correct diagnosis is often long and includes more than one diagnostic modality such as plain film radiography, scintigraphy, video fluoroscopy,
arthrography, arthroscopy, MRI, and/or CT. These imaging techniques frequently prove to be inadequate or equivocal in determining the source of the pain. In many cases, the source of wrist pain is due to dislocation or luxation of the carpal bones due to ligament disruption. A correct diagnosis in an early stage is important as the chances are then optimal for the most favourable therapeutic option, i.e., restoring normal anatomy.

In radiology wrist disorders are quantitatively analyzed by the determination of angles between the carpal bones (Fig. 5) from radiographic lateral projections [10–13]. Due to the 2-D nature of the measurements, however, the angles show large variability and their diagnostic use is limited. In more recent in vivo studies, 3-D information about the relative position and orientation of the carpal bones was analyzed using spatial marker techniques [14] and surface registration [15,16]. To this end, the wrist was imaged in different postures by 3-D CT. The diagnostic application of these techniques is promising since estimates of all six degrees of freedom can be determined from the movement of the carpal bones (three translation and three rotation parameters).

The aim of this study was to develop image processing techniques for 3-D MR and CT images of the wrist joint to aid diagnosis in case of wrist complaints. Segmentation and registration techniques were developed to obtain 3-D geometric descriptions of the bones in the wrist, and to determine the
four-dimensional (4-D) kinematics of this joint, respectively. My goal was to quantify and visualize the movements of the carpal bones by determination of their relative translation and rotation. Two movements of the hand were studied: ulnar–radial deviation and flexion–extension. In order to acquire the kinematic data, the hand was placed in a large number of different postures (typically 15–20). At each posture, which represented an intermediate position between, for example flexion to extension, the wrist was imaged by MRI or CT.

1.4 Image Segmentation

The segmentation of medical images is complicated by the complexity and variability of anatomical shapes. Moreover boundaries between different anatomical structures are often indistinct or even absent because of lack of contrast between these structures, or due to sampling artifacts, spatial aliasing, partial volume effects and noise. This may seriously hamper the accuracy of the segmentation.

Traditional image-processing techniques consider mostly only local information and may produce incorrect boundaries. Therefore, these model-free techniques usually require a high degree of interaction by an expert to obtain a correct boundary. Another shortcoming is that they generally produce roughly discretized segmentations which do not reflect the smooth anatomical shape of the structures which, as a consequence, may counteract the accuracy of particular measurements. Instead, geometric representations are required to capture the idea of 2- and 3-D shape in order to produce realistic visual output and to provide quantitative power [17].

I have chosen deformable contour and surface models for image segmentation of the wrist since they employ geometric representations that allow a large variety of shapes [9]. In addition, their physics-based, intuitive features impose constraints on how their shape may vary over space and time. In physical terms, the deformable models are considered as elastic bodies which naturally respond to externally applied forces and constraints. Deformable curve-, surface-, and solid models were introduced in computer vision and graphics by Terzopoulos et al. [18] in the 80s. Kass et al. [19] introduced planar deformable curves generally known as snakes in the area of image analysis. The mathematical foundation of snakes draws from the theory of optimal approximation involving functionals which represent the energetic state of the curve. The functional is composed of external and internal energetic contributions which give rise to external and internal deformation forces. External image forces are derived from a potential energy function associated with the
Figure 6: A potential energy image $P$ (middle) is obtained from a region of interest in a MR image slice (left) by Gaussian gradient filtering. The valleys of the potential surface (right) represent the image features (contours) to be detected.

image (Fig. 6). The minima of the potential surface represent the image features to be detected. The curve will descend into the valleys of the potential surface by attraction of the image forces. Internal forces control the tension and bending of the curve during this descent. When internal and external forces balance each other, the curve is at equilibrium and is considered as an optimal estimate of the true boundary. The snake paradigm of Kass et al. was followed by many other authors and has led to a whole breed of different deformable curves, surfaces, and solid models which can be used for shape recovery and tracking in 2- and 3-D image data. For an overview of these models applied in the field of medical image analysis, we refer to the survey of McInerney and Terzopoulos [9]. Because all existing models caused some problems when they were applied to the wrist data a 2-D deformable contour model was developed to track the carpal bone contours in stacked MR images and a 3-D triangulated deformable surface model to detect the boundary in 3-D MR and CT images of the wrist.

1.5 MR and CT Imaging of the Wrist

In this study MR and CT techniques were used to image the wrist. From a diagnostic point of view, MRI is generally better suited to investigate wrist pathologies than CT, since it can depict soft tissue in high contrast. For quantitative image analysis of the wrist, MRI has some disadvantages like chemical shift artifacts and image distortion due to magnetic field inhomogeneities. In addition, relatively long acquisition times (6–10 min) are required until now to obtain sufficiently low-noise, detailed images in 3-D. CT imaging provides higher resolution, low noise, undistorted 3-D images in a shorter acquisition time (1.5–2 min).
The main disadvantage of CT is that the total number of 3-D images which can be recorded of a single subject is limited due to the necessity to keep the exposure due to x-rays within acceptable limits. In order to derive a ‘continuous’ description of the movement of the carpal bones a large number of 3-D images is required. The problem of excessive x-ray exposure was circumvented by using low-dose CT-scans.

I will explain the basic MR and CT imaging principles in the following sections. For detailed introduction of both techniques, I refer to a textbook on medical imaging, e.g. [20].

### 1.5.1 Magnetic resonance imaging (MRI)

A MR scanner (Fig. 7) consists of a magnet with a high field strength, usually a large super-conducting coil which surrounds the body and which is cooled by liquid helium. Inside the coil, a magnetic field of 0.5–1.5 T is present, about a factor 10,000–30,000 larger than the field strength of the earth. This field has the effect of aligning the protons of the hydrogen atoms.\(^1\)

Normally, the magnetic moments or spins of the protons will be oriented randomly due to motions produced by thermal energy. Therefore, the macroscopic magnetic moment (\(M\)) is zero. When a static magnetic field \(B_0\) is applied, such as the one present in the MR scanner, a slight excess of spins originate with orientations in the field direction. Consequently, a magnetization component in this direction is induced. The spins experience a torque, which causes them to rotate around the axis of the applied field with a precise

\(^1\)A number of other nuclei will be influenced as well. Because of their abundance in the human body, however, hydrogen atoms serve as the basis for MRI scanning.
frequency, called Larmor frequency (Fig. 8). The transverse magnetization component is zero since the individually precessing spins are still randomly oriented with respect to their phase. By application of a radio frequency (RF) pulse matching the Larmor frequency of the protons the spins are excited. The RF pulse induces an oscillating magnetic field \((B_1)\) that gives rise to a transverse rotating precession axis. As a result, the magnetization vector is rotated by an angle \(\alpha\), called flip angle, which depends on the magnitude and duration of the RF pulse. A net transverse magnetization component \((M_{xy})\) can be measured because it induces a current in a receiver coil (Fig. 9). After excitation, the spins return to equilibrium (relaxation), dephasing by mutual interactions and losing energy to the surroundings (‘lattice’) by emitting electromagnetic radiation. The spin-lattice relaxation is characterized by the longitudinal relaxation time \(T_1\), and the spin-spin relaxation by the transverse relaxation time \(T_2\). Both relaxation times are specific for the molecular structure and composition, physical state (i.e., liquid or solid) and temperature of the tissue.

Spatially localized NMR is obtained by encoding techniques. Three magnetic field gradients are used for this purpose (Fig 10). To create an image of a slice of human tissue, a longitudinal gradient \((G_z)\) is used for slice selection. Two orthogonal transverse gradients are used for phase \((G_y)\) and frequency encoding \((G_x)\) of the signal, originating from the excited slice after application of a RF pulse. By repeated, successive application of the transverse gradients, a 2-D Fourier decomposition is obtained by sampling the induced signal of the receiver coil. In this way, the frequency- or \(k\)-space is filled. When the amplitudes of the signals at all frequencies are collected, an inverse Fourier transform results in the anatomical image. A 3-D volume is imaged by using \(G_z\) for longitudinal phase encoding instead of slice selection. In that case the whole volume is excited by the RF pulse and a 3-D inverse Fourier transform is applied to create the 3-D image.

For an image to have diagnostic utility, there must be contrast between the MR signal of different tissue types. The contrast would be limited if the signal only would depend on proton density. Other parameters, in particular \(T_1\) and \(T_2\) relaxation times, influence the contrast of the tissue to a larger extent than does proton density, however. The relative contribution of both relaxation times can be influenced by adjusting the RF pulses, and the gradients applied, as well as the timing of the data acquisition. This is established by setting var-

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2 This is a classical explanation of the magnetic phenomenon. For a correct description, however, quantum theory is required.

3 Note that the three gradients \(G_x\), \(G_y\), and \(G_z\) do not necessarily have to coincide with the directions of the corresponding gradients in hardware, indicated in Fig. 10. For the preferred slice orientation, the direction of the gradients can be adjusted by linear combinations of the hardware gradients.
**Figure 8:** Precession of a spin $\mu$ about the axis of the applied magnetic field $B_0$.

**Figure 9:** The magnetization vector ($M$) before (left) and after application of a RF pulse (right). When the magnetic field is parallel with the $z$-axis, all spins are aligned and the net transverse magnetization ($M_{xy}$) is zero in the $xy$-plane. During application of a RF pulse matching the Larmor frequency of the nuclei, an oscillating magnetic field ($B_1$) is induced. This gives rise to a transverse rotating precession axis. As a result, the magnetization vector ($M$) is tilted over a flip angle ($\alpha$) and a net transverse magnetization component ($M_{xy}$) can be measured.
ious parameters in a MR pulse sequence, like the pulse repetition time (TR), the time between excitation and maximum echo signal (TE) and the flip angle (α). An example of a 3-D image of a wrist obtained by the so-called 3-D fast low angle shot (FLASH) sequence used in this study is shown in Fig. 26 (top), p. 51.

1.5.2 X-ray computed tomography (CT)

A computed tomography system (Fig. 11) consists of a collimated x-ray tube and an array of detectors which are placed at opposite sides on a circular gantry. The part of the body to be imaged – the wrist for instance – is placed in the centre of the CT scanner. Subsequently, cross-sectional images of the tissues within the body are obtained by revolving the gantry around the body. During a single revolution, the x-ray transmission through a slice of the body is measured along lines from many angles and directions using a fan beam of x-rays and a number of detectors (Fig. 12). When all data for the reconstruction of one image slice are collected, the body is shifted through the gantry to expose the next slice.

A cross-sectional image of each radio-graphed slice is reconstructed from the transmission measurements made in the slice (Fig. 13). The measured intensity at a single detector cell depends on the attenuation of the intensity along a pencil beam, which gives rise to a line integral. Reconstruction of the image from these line integrals is equivalent to the determination of the spatial distribution of the linear attenuation coefficient $\mu(x, y)$. In general $\mu(x, y)$

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*I consider a third-generation CT system here. Later generations use a fixed detector ring.*
Figure 11: The spiral CT imaging system that was used to image the wrist (Elscint CT-Twin/Flash, Elscint Inc., Hackensack, NJ).

Figure 12: X-ray transmission through a slice of the body is measured along lines from many directions using a fan beam of x-rays coupled to an array of detectors.
Figure 13: Reconstruction of a CT image slice of the wrist. The linear attenuation coefficient is calculated from parallel intensity projections along pencil beams through the specimen. The projected intensity is shown logarithmically for three projection angles.
is calculated by convolution and back-projection techniques. After reconstruction, the 2-D images can be stacked to constitute a 3-D image.

The technique as described above is known as conventional CT. In this study another CT technique, spiral CT, was used. In spiral CT the body is moved continuously through the revolving gantry instead of being shifted stepwise. As a result, the pencil beams used for reconstructing a single slice are not co-planar as in the conventional case, but distributed along a helix, which necessitates additional interpolation. The main advantage of the spiral technique is that images can be obtained at higher speed, and can be reconstructed at smaller increments than the slice spacing, which is especially useful for high-resolution 3-D imaging. In the race for higher acquisition speed multi-slice CT technology is currently introduced in radiology to scan simultaneously multiple slices. An example of a 3-D image of a wrist obtained by spiral CT is shown in Fig. 26 (bottom), p. 51.

1.6 Overview of this Thesis

The 2-D deformable contour model that was used to track the contours of the carpal bones in stacked MR data is treated in Chapter 2. This chapter is an extended version of the article that was published in the IEEE Transactions on Medical Imaging [21].

The 3-D deformable surface model that was used to extract the surfaces describing the carpal bones and the facets of the metacarpal bones, radius, and ulna in 3-D MR and CT images, is treated in Chapter 3. This chapter is an extended version of the manuscript that was submitted to the IEEE Transactions on Medical Imaging [22].

The registration technique used to determine the four-dimensional (4-D) kinematic parameters from a sequence of 3-D CT images of the hand is explained in Chapter 4. This chapter will be published as a research article in Medical Physics [23] and an abstract was presented as a scientific paper at the 85th Scientific Assembly and Annual Meeting of the Radiological Society of North America [24].

The techniques used for the image analysis of the wrist are discussed in Chapter 5. This thesis is completed with a summary in Dutch.