# Colour Doppler Imaging:
## New Applications in Musculoskeletal System Pathology

# Colour Doppler Imaging:
## Nieuwe Toepassingen bij Aandoeningen van het Bewegingsapparaat

**Proefschrift**

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**Prof.Dr. P.W.C. Akkermans M.A.,**

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door

**Hızır Muzaffer Buyruk**
geboren te Istanbul, Turkije

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**to Demir, Aylin and Irene**

**thanks to my parents**

# Contents

## Introduction

In the departments of Rehabilitation, Biomedical Physics and Technology, and Anatomy research and clinics of the musculoskeletal system form the centre of attention. In this field pathologies related to low back pain and hand and wrist are under investigation. In both areas some quantitative assessments for diagnosis, treatment and follow-up can not be done with existing routine techniques.

For normal examination procedures many means are available. Examples of clinical examinations are Laseque; imaging techniques, e.g. X-rays, CT and MRI; functional laboratory techniques e.g. dynamometry and EMG, and special facilities like bone traction.

In the study on low back pain special interest raised for the mechanical properties of the pelvic joints, in particular the sacroiliac (SI) joints. For the assessment of SI joint stiffness no instrumented method was available. In the study on the hand and wrist pathology no method was available to measure non-invasively the function of the tendons. While searching for solutions we discovered that Colour Doppler Imaging (CDI) opened new possibilities in the research of the musculoskeletal system in general, and for our research interests in particular. In the following paragraphs a general introduction is given on the existing routine use of CDI. Further, application on the assessment of SI joint stiffness, and hand and wrist pathologies is introduced in separate sectors which give a brief overview of the chapters of this thesis. The general aim of this study is to extend the application of CDI in the field of the musculoskeletal system.

## a. Doppler Technique

### Diagnostic Ultrasound and Doppler Flow Imaging

Diagnostic ultra-sonography is medical cross-sectional anatomic and flow imaging using pulse-echo ultrasound. It is an interactive process involving the radiologist, patient, transducer, instrument and occasionally radiology technician. Anatomic imaging with ultrasound is accomplished with a pulse-echo technique. Pulses of ultrasound are generated by a transducer and sent into the patient where they produce echoes at organ boundaries and within tissues. These echoes return to the transducer, where they are detected and then imaged on an instrument. The transducer both generates the ultrasound pulses and detects the returning echoes. Ultrasonography requires the knowledge of the location of origin and the strength of the echoes returning from the patient. The ultrasound instrument processes the echo information and generates the appropriate dots, which form the ultrasound image on the display. The brightness of each dot corresponds to the echo strength, producing what is known as a grey-scale image. The location of each dot corresponds to the anatomic location of the echo-generating structure. The positional information is determined by knowing the direction of the pulse when it enters the patient and measuring the time for its echo return to the transducer. From an assumed starting point on the display (usually at the top), the proper location for presenting the echo can be derived, provided the direction in which to travel from that starting point to the appropriate distance is known. With the knowledge of the sound propagation speed, the echo arrival time can be converted to the distance to the structure that produced the echo. Ultrasound instruments use the arrival times of the echoes to locate objects properly in depth. If one pulse of ultrasound is emitted, one series of dots (one line of information or one scan line) is displayed. Therefore, not all of the ultrasound pulse is reflected back from any interface. Rather, a portion of the original pulse will continue on and reflected back from deeper interfaces. Scan formats are commonly limited to two types: linear and sector. In all format types ultrasound pulses be sent through all the portions of the cross section that is to be imaged. Each pulse generates a series of echoes, which results in a series of dots (scan line) on the display. The resulting cross sectional image is made up of many (around 200) of these scan lines. The scan format determines the starting points and path for each pulse used in generating each scan line. These are sometimes called B scans, reflecting the fact that the images are produced by scanning the ultrasound through the imaged cross section (i.e., sending pulses through all regions of the cross section) and converting echo strength into brightness of each represented echo on the display (hence, B-scan, or brightness scan). If an echo-generating structure is moving, the echo will have a different frequency than that of the pulse emitted from the transducer. This

will be displayed in M mode which shows reflector motion in time. The changing frequency will create the Doppler effect, which is put to use in blood flow detection and measurement. The Doppler instrument determines the change in frequency resulting from the motion, converting it to an audible sound and presentation on a display that represents the flow.

**The Doppler effect**

When an observer is moving relative to a wave source, the frequency he measures is different from the emitted frequency. If the source and observer are moving towards each other, the observed frequency is higher than the emitted frequency; if they are moving apart the observed frequency is lower. This simple but important assertion is known as the Doppler effect, after the Austrian physicist Christian Doppler (1803-1853), who first postulated it in a paper given before the Royal Bohemian Society of Learning in 1842 (White 1982), and which was published in the proceedings of that society during the following year (Doppler 1843).

The Doppler effect has also been widely used in terrestrial applications, and it is only relatively recently that it has become important in medicine. Both ultrasound and laser Doppler have found a place for detecting motion within the body, but the ultrasound technique is much more widely used. Doppler ultrasound has been used for many years in medicine (1957) as a tool for monitoring fetal heart rate during labour and for detecting and quantifying blood flow in the heart and a variety of vessels throughout the body. This flow information is presented as an audible sound and visually as a flow-versus-time plot or as a colour-coded two-dimensional presentation.

Doppler instruments respond to moving reflectors or scatterers (usually blood cells in circulation) by detecting Doppler shift. This information is converted to audible sound and to visual displays. Doppler instruments are of four types:

1. Velocity detecting systems- Transmitting microwaves or sound, various types, mostly non-directional audio output, no imaging, very simple hand form to complex big units; subgroups according to their wave propagation are:

    a. Continuous-wave (CW) spectral (1966)- These systems provide motion, and flow information without depth information or selection capability. Spectral analysis provides the information on the distribution of the received frequencies resulting from the distribution of scatterer velocities (speeds and directions encountered).

    b. Pulsed-wave (PW) spectral (1967)- These systems provide, in addition to information from CW Doppler, depth information and the ability to select the depth at which Doppler information is generated

    c. Adaptive pulsed Doppler systems

2. Duplex systems (1974)- Duplex scanners are devices that combine a pulse-echo B-scan system and a Doppler system so that the Doppler shift signal can be recorded from known anatomical locations. There are a number of ways of combining the two modalities, but they all share certain characteristics; the permissible directions for obtaining Doppler information all lie within the scan plane of the pulse-echo imager, and the direction of the Doppler beam at any instant is indicated by a cursor superimposed on the image.
3. Profile detecting systems- PW multichannel Doppler systems
4. Velocity imaging systems
    a. CW Doppler Imaging
    b. PW Doppler Imaging
    c. Real-time two dimensional colour flow mapping (1981)

**Real-time two dimensional colour flow mapping**

Combined systems (Duplex) utilizing dynamic B-scan imaging and PW and CW Doppler are available. Colour-flow systems provide displays of two-dimensional, real time flow superimposed on grey-scale anatomic scans. To a large extent the images of vessels produced by Doppler flow mapping systems and

ultrasound pulse echo systems are complementary. The former provides an image of the flow channel, but not of the vascular wall; the latter provides an image of the healthy vessel wall but not necessarily of the flow channel, as many non-calcified lesions mimic blood with their low acoustic impedance. Thus the combination of these two types of images is potentially of great value.

Axial resolution is determined by spatial pulse length. Lateral resolution is determined by beam width. The beam width perpendicular to the scan plane causes section-thickness artifacts. Apparent resolution close to the transducer is not directly related to the tissue texture, but is a result of interference effects from a distribution of scatterers in the tissue (speckle). Reverberation produces a set of equally spaced artifactual echoes distal to the real reflector. Refraction displaces the echoes laterally. In a mirror-image artifact, objects that are present on one side of a strong reflector are presented on the other side as well. Enhancement results from low-attenuation objects in the sound path. Propagation speed error and refraction can cause objects to be displayed improperly in terms of location or size, or both. Refraction can also cause edge shadowing.

Artifacts with Doppler ultrasound include aliasing, range ambiguity, colour-flow image and Doppler signal mirroring, and spectral trace mirroring. Aliasing is the most common artifact. It occurs when the Doppler shift frequency exceeds one half of the pulse repetition frequency (PRF). It can be reduced or eliminated by increasing the PRF or Doppler angle, using baseline shift, reducing the operating frequency, or using a CW instrument.

Phantoms and test objects provide means for measuring detail resolution, distance accuracy, compensation, sensitivity, and dynamic range of diagnostic instruments. Hydrophones are used to measure the acoustic output of diagnostic instruments.

There have been no independently confirmed, significant bio effects reported in mammalian tissues exposed to focused spatial peak-temporal average (SPTA) intensities of less than 1 W/cm$^2$. No risk with use of diagnostic ultrasound in humans has been identified. Because there is limited specific knowledge, however, a conservative approach is appropriate, viz. diagnostic ultrasound should be performed only with minimum exposure and when medical benefit is expected to be derived from the procedure.

**b. Doppler Measurements in Pelvis**

Low back pain is a common peri- and postpartum complaint which can lead to permanent disabilities. The possible relation between low back pain and the stability of pelvic joints has been shown in earlier. In vivo assessment of the stability of pelvic joints is a problem in clinical medicine. Although there are manual methods to survey the condition of these joints, many of them are unreliable and subjective. In earlier studies, various invasive or non-invasive experimental techniques were employed to assess pelvic joint stiffnesses such as x-ray stereometry with insertion of tantalum balls and measurements with goniometers.

In order to generate displacement in SI joints, force application to pelvis is a suitable method to use on embalmed human specimens. This method is not practicable to apply on living people. Imaging techniques like X-rays, computerised tomography or magnetic resonance imaging are very useful for the diagnosis of visible pathologies causing secondary SI instability such as tumours or cysts. However, the movements in the joints are relatively small to display with the help of these imaging techniques.

Dynamic examination with application of vibrations were used in vivo on bones and soft tissues. In traumatology, the healing of the fractured bone was tested with the signals which derived from applied vibrations, and received by accelerometers. The use of a conventional grey-scale ultrasound combined with Doppler, which is not older than a decade, is one of the major developments in medical diagnostic methods. Unfortunately, the use of this apparatus has been restricted to blood flow examinations, even the hardware and software built in the machine. The reaction of the coloured Doppler apparatus to vibrated soft tissues was studied by different groups.

The present thesis deals with the introduction of a new method for the in vivo assessment of SI joint stiffness. The idea is that the iliac bones and the inter-adjacent sacrum can be considered a mass-spring system. The stiffness of the elastic connection of ilium and sacrum can be measured by excitation

of ilium with a vibration which is transmitted by the connection without energy loss in case of stiff connection and with large energy loss in case of loose connection. To validate this method four studies were designed: 1) experiments on a plastic model; 2) experiments on embalmed human specimens; 3) tests on healthy subjects; and, 4) comparative study of healthy controls.

**1) experiments on a plastic model:** The aim of this study is to demonstrate the proportional relation between joint stiffness and transmission of vibrations through the SI joints.

**2) experiments on embalmed human specimens:** The aim of this study was to determine the validity, objectivity and reproducibility of stiffness measurements on SI joints of human specimens.
**3) tests on healthy subjects:** In this study the following questions will be answered; can the method be applied in vivo?; what is the reproducibility in vivo?; and what is the range of stiffness that occurs in healthy volunteers?
**4) study on peripartum pelvic pain patients and healthy controls:** The research question of the present study was: are SI joint stiffness levels of PPPP patients different from those of healthy subjects. An attempt was made to find an answer to this question by comparing the measurement results from PPPP patients with a healthy control group.

**c. Doppler Measurements in Hand and Wrist**
Apart from real-time ultrasonography, most of the techniques used to examine the musculoskeletal system (X-ray, CT and MRI) produce mainly static images. Although Colour Doppler Imaging (CDI) is generally used in the examination of blood flow, it may offer the opportunity to study the dynamic characteristics of tendons and muscles.

Six studies on different possibilities of CDI are presented in this thesis: 1) the functional anatomy of hand and wrist of healthy subjects; 2) the pathological conditions of unilateral flexor and extensor tendons of the hand compared with the healthy side; 3) the measurement of tendon displacement velocity; 4) the measurement of tendon displacement excursions in healthy subjects and patients with unilateral hand flexor pathologies; 5) the calibration of tendon displacement measurements with CDI on a phantom model; and 6) the in vitro calibration of Doppler displacement measurements on human specimen arms.
**1) the functional anatomy of hand and wrist of healthy subjects:** The aim of this study was to determine the applicability of CDI in the dynamic examination of the anatomical structures of the wrist and hand region.
**2) the pathological conditions of tendons in hand and wrist:** The purpose of the present study is to determine the applicability of CDI on the normal and the pathological conditions of the flexor and extensor tendons.
**3) the measurement of tendon displacement velocity:** The purpose of the present study is to show whether the non-invasive quantification of tendon velocity is possible with CDI and to test the reproducibility of tendon velocity measurements which will be executed on healthy volunteers.
**4) the measurement of tendon displacement excursions:** The aim of the present study is to assess the applicability of tendon displacement measurement by means of CDI on healthy volunteers and patients with hand pathologies and to present the preliminary results. To obtain absolute values the calibrations on the phantom model are used.
**5) the calibration of tendon displacement measurements with CDI on a phantom model:** The aim of this study is to calibrate our Colour Doppler displacement measurements with precise digital displacement measurements; and to develop and to test the necessary filters and correction constants to improve the preciseness.
**6) the in vitro calibration of Doppler displacement measurements on human specimen arm:** The purpose of this study is to calibrate tendon displacement measurements in human specimen arm on different flexor tendons and different tendon excursions.

# Chapter 1

This chapter is mainly based on the following publication

HM Buyruk, CJ Snijders, WPJ Holland, A Vleeming, HJ Stam. Assessment of Sacroiliac Joint Mobility by Dynamic Testing of a Pelvis Model (accepted J. Clinical Biomechanics) 1996.

# Assessment of Sacroiliac Joint Mobility by Dynamic Testing of a Pelvis Model

## Chapter 1

**Abstract**
This study investigates the behaviour of mass-spring systems of a pelvis model under vibrations, and assesses the effect of different artificially created stiffnesses on bilateral sacroiliac (SI) joints. The aim of this study is to demonstrate the proportional relation between joint stiffness and transmission of vibrations through the SI joints. A mechanical model of the pelvis was developed which represents the shape of the pelvic bones, the properties of the joints and major ligaments. While simulating a person lying in supine position, unilateral excitation of the pelvis at the anterior superior iliac spine generated vibrations which were received at the dorsal surface of the homo-lateral ilium and the sacrum by two accelerometers. Four different stiffness levels of the SI joints were combined as ten different situations, by means of metal plates, adjustable screws and by tensions applied on artificial ligaments. The occurrence of resonance peaks in the ten simulated stability situations were evaluated in the frequency range between 25 and 600 Hz. The results showed that different SI joint stiffness levels in a mechanical pelvis model result in different resonance frequencies and transmission of vibrations across the SI joints. The transmission of vibrations through the SI joint was proportional to joint stiffness. The stiffness of the contralateral joint influenced the dynamics of the joint on the measured side.

**Relevance**
The approach of joint stiffness measurement adopted in this paper gives an insight in the transmission of vibrations in a complex multiple mass-spring system which resembles a human pelvis. The experiments with controlled stiffnesses in this study are relevant for the understanding and validation of the Doppler Imaging of Vibrations (DIV) which has been recently introduced for the in vivo assessment of SI joint stiffnesses.

**Keywords**
Pelvis, Biomechanics, Mobility, Vibration, Sacroiliac joint, Model

### Introduction

Assessment of the sacroiliac (SI) joint mobility is an unsolved clinical problem. Objective testing of the sacroiliac joint mobility is at present restricted to invasive methods. Non-invasive test methods, such as pain provocation and mobility evaluation tests, are subjective and accuracy of the results is doubtful[1,2].

The need for a reliable and non-invasive assessment of SI joint stiffness initiated the question whether a measuring technique could be developed that enables quantification of SI joint mobility. A possible method could be the load deformation test, which is feasible in embalmed human specimens by application of force to the bones[3,4]. In living persons, however, problems arise because of the very small SI joint movements in comparison to disturbing influences like skin deformation and restricted load level. Applying vibrations, as used for the examination of bone stiffness and fracture healing, could be a useful alternative[5,6,7,8,9,10,11].

According to existing literature in this field, Colour Doppler Imaging (CDI) has been used in combination with vibrations to assess the differences in stiffness of soft tissues[12]. We performed studies in which vibrations and CDI were used to assess the stiffness of SI joints of human specimens[13], healthy subjects[14] and peripartum pelvic pain patients(PPPP)[15] . This technique was called Doppler Imaging of Vibrations (DIV). It appeared that vibrations were easy to apply and that it is feasible to work with forces of low level[16]. In human specimen studies it was possible to discriminate different stiffnesses of SI joints. The study with healthy subjects showed that the measurements with DIV are reproducible and reliable. Furthermore, in this study non-symptomatic healthy subjects presented various stiffness levels of SI joints. In the comparative study with PPPP patients and healthy subjects, both groups showed loose as well as stiff

SI joints. However, the patient group showed a significant difference between stiffnesses on the left and the right-hand side.

The promising results from the studies with DIV led us to investigate the behaviour of a pelvis under vibration. A theoretical model would be too complex for the description of the dynamic behaviours of the different pelvic structures, which is shown in 1.1. The existing transmission systems in a vibrated pelvis are simplified as lumped parameter model in figure 1.1[13,14,15]. The masses, M1 and M3 represent the iliac bones and M2 represents the sacrum. The damped mass-spring systems indicated with T2 and T3 represent the right and left SI joints, respectively, and T4 is the symphysis pubis. T1 represents the contact with the vibrator (V), which acts on the anterior superior iliac spine. T5 is the contact of the pelvis with the laboratory frame (G). In this study we investigate the behaviour of a pelvis model under vibrations with the mass-spring systems mentioned above, and assess the effect of different artificially created stiffnesses of T2 and T3. The aim of this study is to demonstrate the proportional relation between joint stiffness and transmission of vibrations through the SI joints.

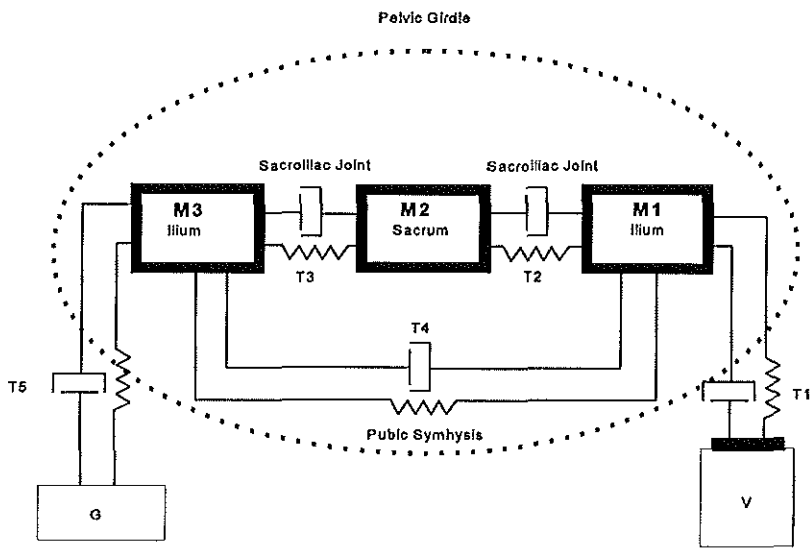


**Figure 1.1.** Model of the sacroiliac joints (T2 and T3) in the vibrating pelvis; G rigid laboratory frame, V excitator **M (1..3)** masses, **T (1..5)** transmission systems with springs and dampers.

### Materials and Methods

The mechanical model designed for this study represents schematically the geometry of a human pelvis (Figure1.2). The exploded view of the model is shown in figure 1.3. The dimensions, type of materials, and the simulated anatomical structure are given in table 1.1. The size of the model is chosen in ratio to a macerated female human pelvis with a width of the iliac crests of 30 cm. In order to simulate the weight of the body and the legs three brass rods were placed in the model, one on the cranial and two on the caudal side. Three metal masses were connected to the rods, one of 8 kg on the cranial and two of 5 kg on the caudal rod.

In the experiments an electromagnetic vibrator (Derritron VP4/TA120) equipped with control unit was used. This can generate vibrations between 1.4-20000 Hz with adjustable power output levels. For the quantification of vibration amplitudes, two piezoplastic accelerometers with built-in amplifiers (Hakuto Super G-1A; 0.5-600 Hz) were used[17,18]. Vibrations were measured at the dorsal side of the structures representing the right ilium (A1) and the sacrum (A2). Data derived from the accelerometers were processed by a digital computing oscilloscope (Le Croy 9410 150 MHz). Three digital displacement metres (Mitutoyo IDU25E) were used to measure the displacement of the sacrum under load.

The mechanical properties of SI joints and pubic symphysis were simulated. Four different stiffness

levels of the SI joints were combined unilaterally and bilaterally resulting in ten situations (P1..P10) (Table 1.2). These four levels were unstiffness (hyper-mobility), medium stiffness (muscle and ligament active stability), stiffness (hypo-mobility because of irregular joint surfaces) and absolute stiffness (ankylosis). At every stiffness level the primary stability of the SI joints was simulated with polypropylene strips (Table 1.1; nr. 5, 11) and cylinder formed rubber (Table 1.1; nr. 12), which represented anterior, posterior and intra-articular SI ligaments in situation P7, respectively. Secondary stabilities, such as increasing tension on the sacrotuberal ligament during contraction of the biceps femoris muscle[19], were simulated with aramide ropes (Table 1.1; nr. 9) in situations P3, P6 and P10. Stability created by irregular joint surfaces was simulated with two iron screws (Table 1.1; nr. 1) in situations P2, P5 and P9. The condition of ankylosis in the SI joints was simulated by replacing both polyamide strips with the same size 1 mm thick steel plates (Table 1.1; nr. 5, 11) at the ventral and dorsal side of the joints in situations P1, P4 and P8.

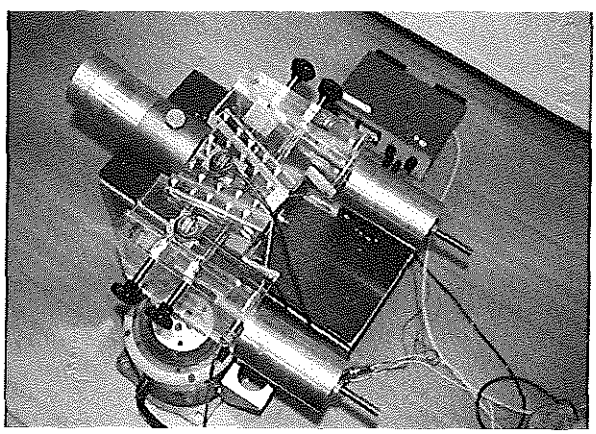

**Figure 1.2.** Mechanical model.

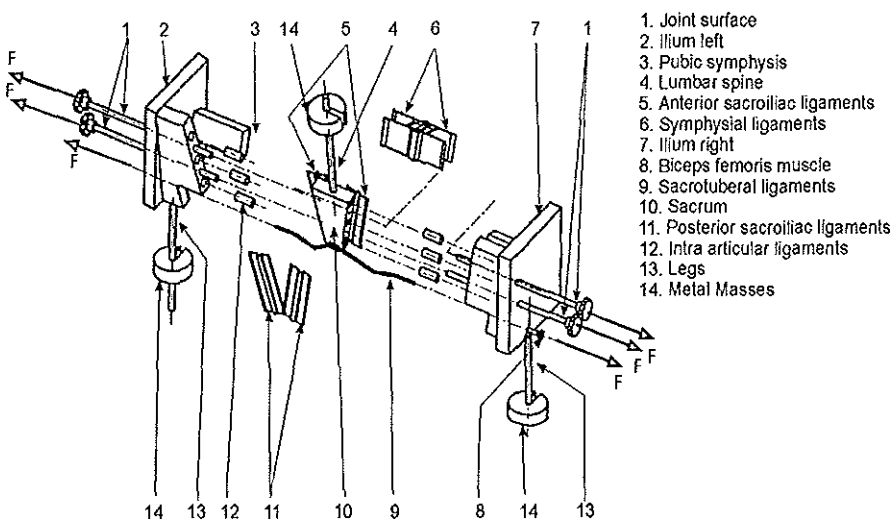


**Figure 1. 3.** Exploded view of the model

Table 1.1. List of simulated anatomical structures, type of materials and their measures

| Nr. | Anatomic Structure | Material | Measures (mm) | Elasticity (E-modulus) |
|---|---|---|---|---|
| 1 | Irregular joint surface | Steel 360 (screws) | M10x250 | 2.0 E5 MPa |
| 2 | Ilium left | Polymethylmetacrylate | 200x200x30 | - |
| 3 | Symphysis | Polymethylmetacrylate | 240x50x30 | - |
| 4 | Vertebral column | Steel 360 (screws) | M10x200 | 2.0 E5 MPa |
| 5 | Anterior sacroiliac ligaments | Polypropylene | 25x110x1.5 | 1100 MPa |
| 6 | Symphysial ligaments | Polypropylene | 48x10x1.5 | 1100 MPa |
| 7 | Ilium Right | Polymethylmetacrylate | 20x20x3 | - |
| 8 | M. biceps femoris | Brass (screws) | M10x50 | - |
| 9 | Sacrotuberal ligaments | Aramide fibres | ⊖ 3x120 (2x) | 1.3 E5 MPa |
| 10 | Sacrum | Polymetylmetacrylate | 130x(90x40)x50 | - |
| 11 | Posterior sacroiliac ligaments | Polypropylene | 25x110x1.5 | 1100 MPa |
| 12 | Intra-articular ligaments | Butyl rubber | ⊖ 15x40 | 30 Shore |
| 13 | Legs | Steel 360 (screws) | M 10x200 | 2.0 E5 MPa |
| 14 | Weights | Brass | ⊖ 80x26 | - |

The measurements presented in this study only refer to the right-hand side of the model in which transmission system T2 represents the right SI joint (Figure 1.1). The load deformation values of T2 in ten different situations were measured by bidirectional static application of a load of 100 N on a lever arm of 10 cm proximal to the sacrum. The sagittal and transversal rotation values are given in degrees in table 1.2.

The motion pattern of the vibrating pelvis was observed with the help of a stroboscope. The acceleration amplitudes at both sides of T2 (A2 sacrum and A1 right ilium) were recorded every 25 Hz between 25 and 600 Hz, and the ratios (A2/A1) were calculated. These ratios are plotted as three different graphics (Figures 1.4, 1.5 and 1.6). The reproducibility of these measurements was tested by means of four repetitions on both sides of the symmetric model.

**Results**

Observations with the stroboscope displayed a constant but complex circular motion pattern with many directional components. The total bidirectional sacral rotation decreased, as expected especially in the sagittal plane, with the application of different means of stabilization (Table 1.2). Bilateral metal plate application to the pelvis model (P4) created the highest SI joint stiffness while no stabilization at both SI joints (P7) showed the lowest.

In agreement with the arrangement of a damped mass-spring system (Figure 1.1) , the model displayed resonance frequencies. The repeated recordings showed consistency in ratios and resonance frequencies. The response to excitations at different frequencies are shown in figures 1.4, 1.5 and 1.6. The resonance frequencies and the corresponding ratios for every condition are given in table 1.3. In figure 1.4, A2/A1 ratios are plotted during three stabilizations of T2 in situations P1, P2 and P3. Situations P1, P2 and P3 displayed two resonance peaks in the frequency range between 25 and 600 Hz with different ratios (Table 1.3). None of the peaks were more than 2.5. Figure 1.5 illustrates A2/A1 ratios obtained with the same stiffnesses levels of T2 and T3 in situations P4, P5, P6 and P7. In conditions P4, P5 and P6, two resonance peaks occurred (Table 1.3 and Figure 1.5). The second peaks from P4 and P5 were with ratios 4.3 and 3.4. The stiffness level P7 without any extra stabilization for both sides displayed only one resonance peak (Table 1.3 and Figure 1.5). With different stiffness levels applied to the left SI joint (P8, P9 and P10), only one peak occurred in this frequency range (Figure 1.6 and Table 1.2). The last four stiffness situations (P7, P8, P9 and P10) showed rather low peaks i.e. ratios smaller than 2.

The model displayed no resonance in some frequency intervals for example between 225 and 350 Hz. In the oscillation frequencies without resonance, the change in vibration transmission between different

stiffness levels is as follows. The ratios (Figure 1.4) decreased from P1 to P3 (P1>P2>P3). In the same stiffness levels for both sides it was observed that P4, P5 and P6 (Figure 1.5) decrease in ratios with the same order (P4>P5>P6). In comparison to all other conditions P7, P8, P9 and P10 (Figure 1.6) displayed the lowest ratios.

**Table 1.2.** Ten different sacroiliac joint stabilization combinations (P1-P10); metal plates simulate ankylosis; screws simulate irregular joint surface; rope simulate tension on sacrotuberal ligament; and total sacral transversal and sagittal rotation under bidirectional load.

| Situations | Extra Stabilization | | Total sacral rotation (degrees) | |
|---|---|---|---|---|
| | Left SI | Right SI | transversal | sagittal |
| P1 | (-) | 2 metal plates | 0.01 | 0.22 |
| P2 | (-) | 2 screws | 0.25 | 1.96 |
| P3 | (-) | 1 rope | 0.45 | 4.79 |
| P4 | 2 metal plates | 2 metal plates | 0.01 | 0.12 |
| P5 | 2 screws | 2 screws | 0.01 | 1.2 |
| P6 | 1 rope | 1 rope | 0.07 | 3.41 |
| P7 | (-) | (-) | 0.57 | 5.92 |
| P8 | 2 metal plates | (-) | 0.01 | 0.26 |
| P9 | 2 screws | (-) | 0.02 | 2.37 |
| P10 | 1 rope | (-) | 0.88 | 5.18 |

**Discussion**

A non-invasive objective method to assess the mobility of the SI joints does not exist in medicine[1,2]. We introduced an instrumented method DIV, using vibrations and CDI, which was addressed in the literature by our group[13,14,15]. In the present study an attempt was made to demonstrate the proportional relation between the joint stiffness and the transmission of vibrations through the SI joint. A mechanical model was designed which resembles the general geometry and mechanics of the pelvis. With this geometry and material of the model it was possible to reproduce various stiffness levels of the pelvic joints which were changed artificially several times. This control was impossible with human specimens[13]. The stroboscope observations and resonance frequency patterns of the vibrated pelvis indicated that the deformation behaviour of the system could not be formulated mathematically.

To meet with the studies on humans[18,19], the excitation site was chosen at the part of the model which represents the anterior superior iliac spine. The weight of the masses connected to the model was chosen in agreement with experiments on human specimens[13]. Although the amount of the weight is important for the experiments on the pelvis model because of the resonance frequencies, it was not critical for the DIV experiments. There are two important reasons for this conclusion; first we did not observe resonance in vivo; second the range of human weight is wide. Based on the results shown in figure 1.4, 1.5 and 1.6 various levels of joint stiffness, as obtained by different means of stabilization, result in a considerable shift of resonance frequencies and differences in ratios of accelerations picked up at the dorsal side of the ilium and sacrum. However, during the in vivo experiments[14,15] there was no evidence of resonance which would be recognised with the appearance of a higher DIV threshold level on the sacrum than on the ilium. During in vivo measurements resonance frequencies should be avoided. Non-appearance of resonance during in vivo experiments can be explained by the vast damping effect of the soft tissues.

Although the data obtained from the mechanical model can not be fully transferable to organic behaviour, they do present comparable results. Low amplitude ratios appearing with low stiffness values on the measured side (Figure 1.4, P1..P3) can be explained by the decrease of vibration transmission. The same decrease of SI joint stiffness on both sides (Figure 1.5, P4..P7) yielded similar results compared to figure 1.5, which means a decrease

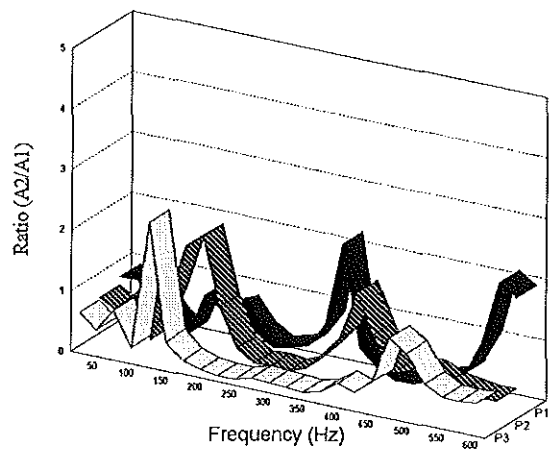


**Figure 1.4.** Graphic displaying resonance frequencies that occur in T2 during situations P1, P2 and P3 when the sacroiliac joint (T2) is stabilized by three different means and T3 is hypermobile.

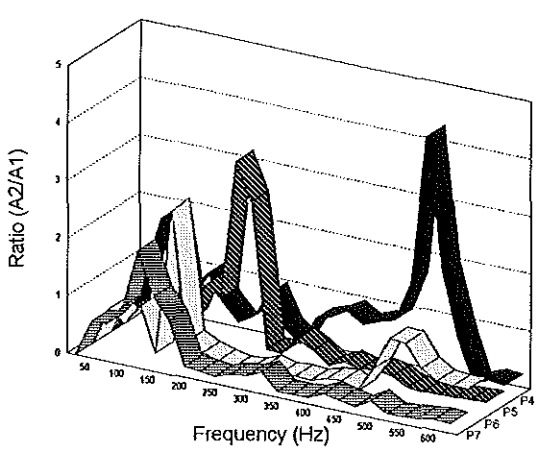


**Figure 1.5.** Graphic displaying resonance frequencies that occur in T2 during situations P4, P5 and P6 when both sacroiliac joints (T2 and T3) are stabilized by three different means; during situation P7 there is no stabilization on both sacroiliac joints and they are hypermobile.

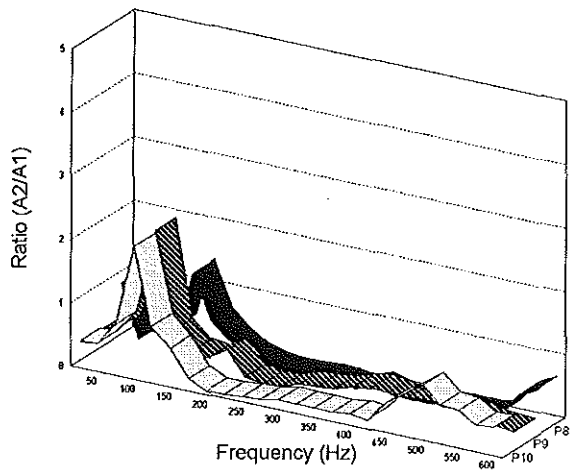


**Figure 1.6.** Graphic displaying resonance frequencies that occur in T2 during situations P8, P9 and P10 when the sacroiliac joint (T3) is stabilized by three different means and T2 is hypermobile.

**Table 1.3.** Resonance frequencies and peaks displayed during excitations.

| Situation | First Resonance Peak | | Second Resonance Peak | |
|---|---|---|---|---|
| | Frequency (Hz) | Ratio (A2/A1) | Frequency (Hz) | Ratio (A2/A1) |
| P1 | 350 | 2.1 | 575 | 2.1 |
| P2 | 175 | 2.1 | 400 | 1.7 |
| P3 | 125 | 2.4 | 475 | 1.4 |
| P4 | 75 | 1.9 | 500 | 4.3 |
| P5 | 125 | 2.0 | 225 | 3.4 |
| P6 | 150 | 2.6 | 500 | 1.1 |
| P7 | 125 | 2.0 | - | - |
| P8 | 125 | 1.3 | - | - |
| P9 | 100 | 2.2 | - | - |
| P10 | 100 | 2.0 | - | - |

in the transmission of vibrations. The decrease of stiffness on the opposite SI joint (Figure 1.6, P8.. P10) increased the transmission of vibrations on the measured side. Similar results were obtained from the study on peripartum pelvic pain patients[15] in which all the subjects with pain showed significant asymmetric SI joint stiffnesses in their pelvis. From both studies it can be concluded that the biomechanics of the pelvic girdle should be considered as a whole and not as a single SI joint.

The damping effect of the skin and the superficial layers are a major disadvantage for the use of accelerometers in vivo. Although accelerometers are the best choice for the model, the damping effect of the

skin in vivo is avoided by DIV. It appeared that during DIV measurements[13,14,15] it is possible to evaluate the amplitude of the vibrating bones on both sides of the SI joints (sacrum and ilium) simultaneously. With the use of Doppler the damping effect of the skin and the calibration problems of the accelerometers are eliminated. The DIV technique allows for the discrimination of hypo-mobile (stiff) joints due to pathology such as ankylosing spondylitis, joints with normal mobility and hyper-mobile (unstiff) joints as a consequence of Ehler-Danlos syndrome or pregnancy.

Even when the excitation method as described here would be used in practice one has to realise the following: The SI joints are an integrated part in the transfer of lumbosacral load to iliac bones and legs[20,21]. This means that clear differences in SI joint mobility do not automatically lead to simple conclusions[15].

## Conclusions

- This study shows that different SI joint stiffness levels in a mechanical pelvis model result in different resonance frequencies and transmission of vibrations across the SI joints;
- The transmission of vibrations through the SI joint is proportional to joint stiffness;
- The stiffness of the contralateral joint influences the dynamics of the joint on the measured side;
- The model displays resonance frequencies. Although it has not been observed earlier, during in vivo measurements attention must be paid to the occurrence of resonance.

## Acknowledgements

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# Chapter 2

This chapter is mainly based on the following publication

HM Buyruk, HJ Stam, CJ Snijders, A Vleeming, JS Laméris, WPJ Holland. The use of colour Doppler imaging for the assessment of sacroiliac joint stiffness: a study on embalmed human pelvises. European J Radiology 1995 21: 112-116.

# The Use of Colour Doppler Imaging for the Assessment of Sacroiliac Joint Stiffness: a study on embalmed human pelvises

## Chapter 2

**Abstract**

**Rationale and Objectives-** The validity and reproducibility of an instrumented dynamic examination method to measure sacroiliac (SI) joint stiffness was tested in vitro.

**Methods-** In this study, four embalmed human female pelvises were excited by a pelvic vibrator. A Colour Doppler Imaging (CDI) scanner was used to image the amplitude of vibrations at different sites of the pelvis. Vibrations were applied to the anterior superior iliac spines unilaterally and were received by CDI all over the ipsilateral SI region. Three different stability conditions were created in the SI joints: no intervention, screwed and ligaments cut. Test results were quantified by taking the minimum threshold levels of the bones. The relative difference of vibration intensity between ipsilateral ilium and sacrum at each stability condition is accepted as the stiffness level for the SI joint.

**Results-** Statistics showed high reproducibility and significant differences between the stability conditions. Dynamic testing based on the use of vibrations provides visible and quantifiable intra- and inter-individual differences between SI joint stiffnesses.

**Conclusions-** This new method is objective and reproducible. Future in vivo application is promising since there are no technical and safety restrictions.

**Keywords**

Pelvis measurements, Doppler studies

### Introduction

Low-back pain is a common disorder which can lead to permanent disabilities; the different stiffness levels occurring in the sacroiliac (SI) joints are considered to be a contributing factor[1]. In previous human specimen studies, the increase in SI joint stiffness has been shown by application of a simple pelvic belt[2]. In vivo assessment of the stability of pelvic joints remains a problem in clinical practice. Although there are many test methods to examine the condition of these joints, all are subjective and many are unreliable[3]. The conventional imaging methods provide no information on the stiffness levels of the SI joints.

Various invasive or noninvasive experimental techniques have been employed to assess pelvic joint stiffness, including X-ray stereometry with insertion of tantalum balls[4], and measurements with goniometers[5]. Force application to the pelvis is a suitable method to generate SI joint movement in embalmed human specimens. Because SI joint movements are small (mm), application of traction with the help of an apparatus cannot be used for in vivo diagnostic purposes. Imaging techniques like X-ray, computerized tomography or magnetic resonance imaging are useful for the diagnosis of impairments which may cause secondary SI instability, such as tumours or cysts. These imaging techniques, however, cannot identify in vivo small SI joint movements.

A possible dynamic method characterized by excitation of vibrations and measuring the response by using accelerometers was previously studied by our group[6] to explore its applicability in measuring joint stiffness on an artificial structure which mimicked pelvic geometry. The mechanical behaviour of this physical model, made of plastic, could be described by a multiple mass-spring system (Figure 2.1) which simulates a patient in supine position. It appeared that the use of vibrations in a safe frequency range, i.e. without resonance, can accurately discriminate stiffness levels of SI

joints. In vivo application of vibrations has been reported in other studies on bones and soft tissues[7-9]. In traumatology, the healing stage of fractured bones was tested with the signals generated by applied vibrations, and received by accelerometers.

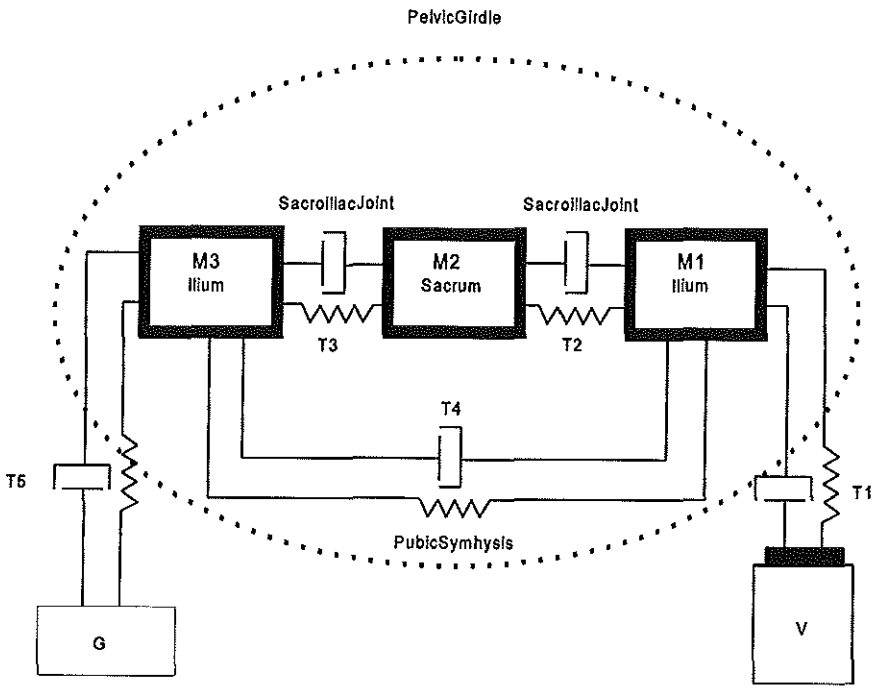


**Figure 2.1.** Model of a pelvis in supine position; T1 to T5 visco-elastic systems, M1 to M3 masses, G ground and V vibrator.

Colour Doppler Imaging (CDI), the combination of conventional grey scale imaging and colour Doppler, is one of the major recent developments in medical imaging. The clinical use of this apparatus has been restricted to blood flow examinations. The hardware and software in the equipment are only specified for testing the blood flow. The reaction of such an apparatus to vibrated soft tissues was studied earlier by different groups to identify tumours in liver and bladder[10]. The differences between the tissue sonoelasticities appeared to have a direct relationship with tissue stiffness.

Based on the successful testing of the SI joint stiffness using a physical model, the question arises whether a dynamic testing method using vibrations and sonoelasticity will also give repeatable and valid outcomes when applied in humans. The aim of this study was to determine the validity, objectivity and reproducibility of stiffness measurements on SI joints of human specimens.

**Materials and Methods**

Dynamic excitation measurements were bilaterally performed on the SI joints of four embalmed female human specimens; age range 92 to 97 years. Pelvises were resected from L4 to mid femur level, keeping all the superficial layers intact. The pelvic organs and layers were dissected, carefully keeping the muscles and ligaments intact. To mimic the masses of legs and trunk, four kilogram of

metal masses were connected to metal rods which were placed in the medulla of the femur and spinal canal and fixed with epoxy resin.

Excitations were generated by a Derritron VP3 vibrator with a frequency range between 1.4 and 20000 Hz and with variable output possibilities. A CDI scan, Angio Dynograph Philips 1, was used to produce the sonoelasticity images. In all pelvises the vibrations were applied unilaterally to the anterior superior iliac spine. Since the angle between the vibration excitation direction and ultrasound propagation should be $0^0$ for maximum Doppler shift the excitations were received on the skin by a 7.5 MHz. CDI transducer perpendicularly positioned on the dorsal region of the SI joint (Figure 2.2). Before the examination, posterior iliac spines and sacral segments were marked on the skin for orientation of the examination area. The sonoelasticity images were digitally recorded on the video, implemented in the apparatus, for further evaluation.

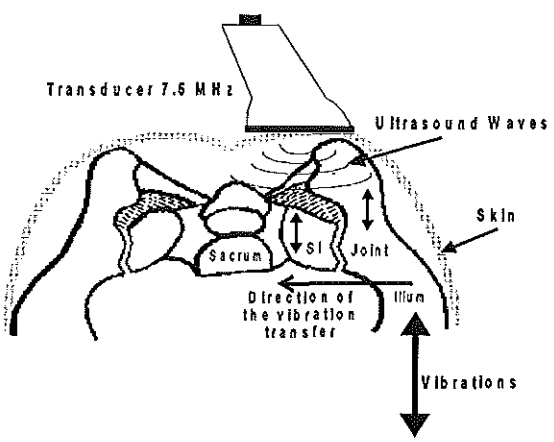


**Figure 2.2.** Experimental set-up showing the direction of excitation of the pelvis resulting in vibrations picked up at both sides of the sacroiliac joint by a Colour Doppler Ultrasound transducer.

Two pelvises were used to gain experience with the technique and to determine the optimal vibration frequencies. We observed the sonoelasticity reactions of the specimens during exposure to different frequencies. Vibrations were applied up to 100 Hz at 10 Hz intervals, between 100 Hz and 500 Hz at 50 Hz intervals, and between 500 Hz and 2000 Hz at 100 Hz intervals. The CDI images obtained from the SI region (Figure 2.3) were evaluated for homogeneity of distribution of Doppler pixel, alternating Doppler threshold levels with different outputs, and superposition of grey scale image with Doppler effect. The rotating threshold button allows to alter the Doppler threshold or the amount power necessary in the signal before it portrayed as vibration. For further experiments we selected 200 Hz.

Two pelvises were used for SI joint stiffness quantification. Sonoelasticity of the sacrum and ilium was measured at three SI joint stiffness conditions: a) no intervention; b) artificially fixed (4 screws through each SI joint, Figure 2.4); c) artificially unstabilized (screws removed and the anterior and interosseous ligaments cut). All SI joints were imaged at least three times to test the repeatability of the sonoelasticity images.

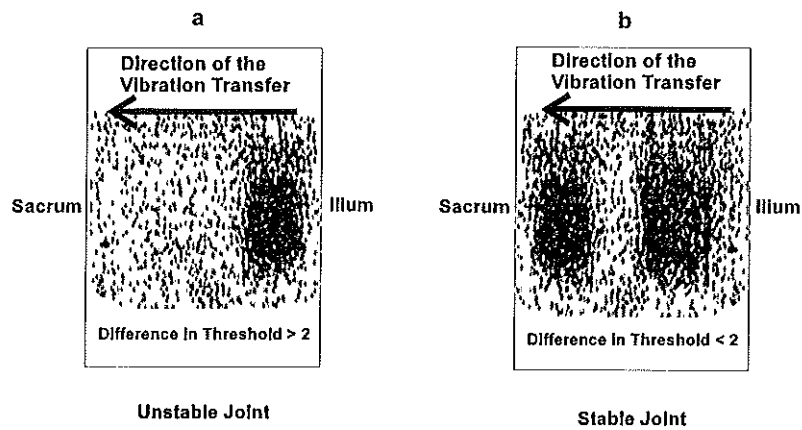


**Figure 2.3.** Doppler densities obtained with different stiffnesses, a) unstable joint; no color Doppler pixels at the sacrum, b) Doppler pixels at both bones; stable joint with high transmission of vibrations.

*Procedure for quantification of the Colour Doppler Image*- The SI region scanned (ilium and sacrum simultaneously) and the minimum threshold value for each bone (the disappearance of the vibrations on the screen) were registered as minimum sonoelasticity levels. The minimum threshold value of the sacrum was subtracted from the value of the ilium. This relative difference between sacrum and ilium represented the power loss of the vibrations over the scanned SI joint. Subsequently, this procedure was repeated at the contralateral SI joint. The results obtained from four SI joints at three different stiffness conditions were statistically assessed with the sign test for reproducibility and significance of sonoelasticity measurements.

**Results**

From the pilot study on two cadavers, it appeared that a vibration of 200 Hz yields optimal information from the images for discrimination of sonoelasticity levels; this same frequency was found in the study on the physical model. In the successive study on two other specimens it was possible to obtain repeatable sonoelasticity images from all of the experiments. The SI joints presented different levels of stiffness during various stability conditions. The unstabilized human specimen pelvises showed a transfer block of vibrations (low stiffness) through SI joints (Figure 2.3a), while the stabilized SI joints showed an increase of vibration transfer (high transfer) (Figure 2.3b). Table 2.1 presents the differences of threshold levels on sacrum and ilium, representing a measure for the stiffness of the SI joint. This measure is not given in a certain unit, but is the result of energy loss over the joint. Repeatability is satisfactory, based on the proportion between the standard deviation and the average of four repetitions. Figure 2.5 shows the clear distinction between stable and unstable joints; from left to right no intervention (a), stabilized with screws (b) and ligaments cut (c). In the no intervention pelvis position (Figure 2.5a) it appeared that both SI joints of specimen 2 show almost no blockage of energy transfer. Fixation of specimen 1 by screws resulted in a dramatic decrease in blockage bilaterally while for the same stabilization procedure no difference was observed in specimen 2 (Figure 2.5b). After removal of the screws and cutting the ligaments (Figure 2.5c) all four SI joints involved showed a clear increase of blockage up to a level above the no intervention position. To assess the significance of these differences the Sign test for left and right separately was applied on the relation between the "screwed" and "ligaments cut" positions. A P value of $<0.001$ was found for both left and right SI joints.

Table 2.1. Threshold differences of the sacroiliac joints with different stiffnesses, mean and standard deviations of the repeated measurements are given in threshold units.

| Pelvis 1 | Unaltered Left | Unaltered Right | Screwed Left | Screwed Right | Ligaments Cut Left | Ligaments Cut Right |
|---|---|---|---|---|---|---|
| 1 | 5 | 4 | 2 | 1 | 5 | 9 |
| 2 | 4 | 6 | 1 | 0 | 7 | 7 |
| 3 | 5 | 5 | 1 | 1 | 7 | 7 |
| 4 | 5 | 4 | 1 | 1 | 6 | 7 |
| Mean | *4.75* | *4.75* | *1.25* | *0.75* | *6.75* | *7.5* |
| SD | *0.5* | *0.95* | *0.5* | *0.5* | *0.95* | *1* |
| Pelvis 2 | | | | | | |
| 1 | 0 | 0 | 0 | 1 | 7 | 5 |
| 2 | 1 | 1 | 0 | 0 | 6 | 6 |
| 3 | 0 | 1 | 1 | 0 | 6 | 6 |
| 4 | 0 | 0 | 0 | 1 | 5 | 6 |
| Mean | *0.25* | *0.5* | *0.25* | *0.5* | *6* | *5.75* |
| SD | *0.5* | *0.57* | *0.5* | *0.57* | *0.81* | *0.5* |

## Discussion

The aim of this study is to develop and validate an objective instrumented method to test the intrinsic pelvic stability. Validation of the method should be performed in vivo. However, at first we were restricted to in vitro testing because the stiffness of an SI joint cannot be artificially changed in vivo in a controlled manner. A "gold standard" has to be obtained in vitro. First it was confirmed that simultaneous sonoelasticity evaluation of the bones of a macerated pelvis (sacrum and ilium) is possible and repeatable with multi-channel pulsed wave CDI. Further, the design of the experimental set-up was directed to the future in vivo application of the new technique. The justification of such application is supported by the measurement results reported here. While long-term exposure to vibrations with low frequency and high amplitude was demonstrated to be harmful to human health[11], high frequencies like 200 Hz with low amplitude, and consequently low energy, have been shown to be harmless and can be used safely in diagnosis[7-9]. With the use of high frequencies the distribution of vibrations is spheric. The advantage of this is that the transfer function measured for the SI joints is not sensitive to the direction of the excitation. For in vivo conditions, the positioning of the patients with respect to the vibrator may not be completely reproducible in prone posture because of differences in body weight and shape. In the earlier study, on a pelvis model made of plastic, a frequency of 200 Hz appeared to be free of resonance.

CDI imaging yields advantages over accelerometers such as; a) simultaneous measurement of two bones (ilium and sacrum) with one sensor, b) no damping influence of the skin, c) insensitive to the thickness of soft tissues, and d) no need for calibration (simultaneous signal reception from different areas is calibrated in the apparatus). The intensity of the vibrations appearing on the screen was affected by the angle between the transducer and the bone surface. This can be explained by the Doppler shift theory in which 0° is the optimal angle. Because we base the measurement on the relative difference of the threshold values found at the sacrum and the ilium this difference did not change, even when the transducer was not perpendicular to the bones.

## Conclusion

Stiffness measurements using sonoelasticity can be applied on human SI joints. Dynamic testing based on the use of vibrations provides visible and quantifiable intra- and inter-individual differences

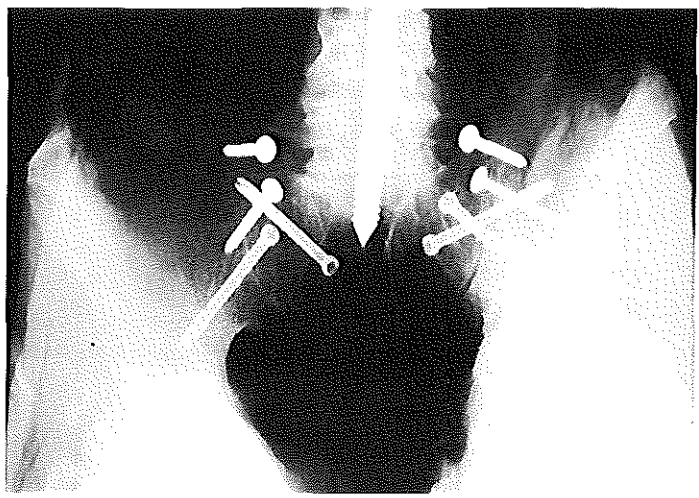

**Figure 2.4.** Stabilization of SI joint bilaterally with four screws on each sides.

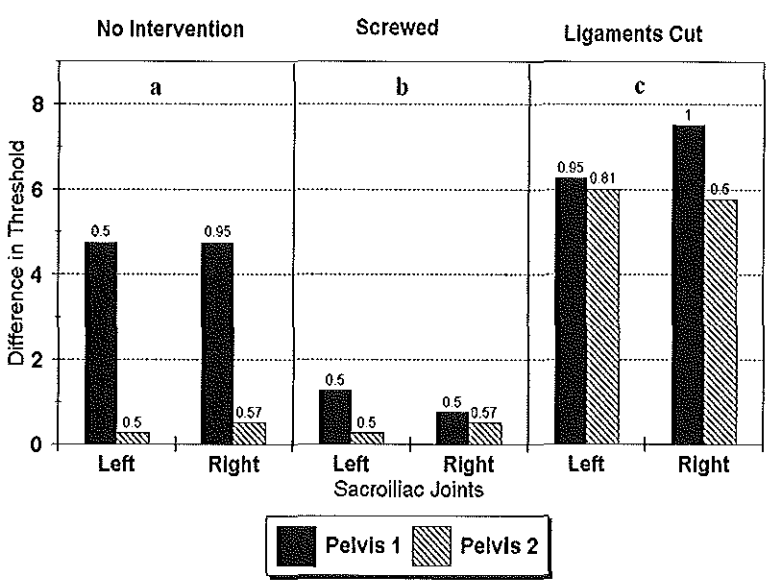


**Figure 2.5.** Levels of the change in Doppler threshold during the energy transfer through the sacroiliac joints at different stiffness condition, a) unaltered, b) screwed, c) ligaments cut are given with standard deviations.

between SI joint stiffnesses. This new method is objective and repeatable, which can be demonstrated with statistical significance by creating different stiffnesses in human specimens. Future in vivo application is promising since there are no technical and safety restrictions.

**Acknowledgements**
Special thanks are extended to Dr. R. Stoeckart, Department of Anatomy, Erasmus University Rotterdam, for his valuable contributions to this project.

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# Chapter 3

This chapter is mainly based on the following publication

HM Buyruk, CJ Snijders, A Vleeming, JS Laméris, WPJ Holland, HJ Stam. The measurements of sacroiliac joint stiffness with colour Doppler Imaging: a study on healthy subjects. European J. Radiology 1995 21: 117-121.

# The Measurements of Sacroiliac Joint Stiffness with Colour Doppler Imaging: a study on healthy subjects

## Chapter 3

**Abstract**

**Rationale and Objectives-** Primary peripartum pelvic and low back pain is a common complaint of females. The etiologic relation between pain and pelvic stability has been shown in previous studies, but at present there is no objective clinical testing method to evaluate pelvic stability.

**Methods-** In this study, a dynamic measurement method using sonoelasticity to assess the sacroiliac (SI) joint stiffness was tested in vivo in 14 healthy female volunteers. With the subjects in supine position vibrations were unilaterally applied to the anterior iliac spine. The vibrations were registered by a Colour Doppler Imaging (CDI) transducer over the ipsilateral SI joint. Since the threshold level of the apparatus has direct relation with the power of the vibrations the intensity of the vibrations (sonoelasticity) on the sacrum and ilium was measured indirectly in threshold units. The differences between the threshold values were accepted as the power loss of vibrations through the SI joint. One-way Analysis of Variance-test and T-test for Paired Samples were applied on the measurement results ($P<0.05$).

**Results-** Statistically, the results showed a satisfactory intra-individual reproducibility and inter-individual variability. There was no significant difference between the data derived from the left SI joint and right SI joint.

**Conclusions-** Based on the promising results on healthy female volunteers, this method will be specifically used in future studies on patients with peripartum pelvic pain.

**Keywords**

Pelvis measurements, Doppler studies

### Introduction

In the literature on low back and pelvic pain many possible causes are reported. It is commonly stated that low back pain is the result of spinal disorders, in particular internal or external disc hernias or lesions of the apophysial joints[1]. Recently a new theory has been introduced[2] . Pelvic instability which is directly related to sacroiliac (SI) joint stiffness[3] might play an important role in the development of low back and pelvic pain. The interest in the mechanics of the SI joints reinforces the need for a reliable measurement method to test joint stiffness. At present clinical tests of SI joint stiffness are done in several ways, such as Patrick or Fabere, Gaenslin and Pelvic Rock tests[4,5]. However, these methods are unreliable and subjective, because the outcome of the examination depends entirely on the experience and skills of the observer, and the test results are not expressed in numeric data. Consequently, a need exists for an in vivo instrumented method which is non-invasive and can be routinely applied in the clinic.

In earlier studies sonoelasticity was described as mechanical palpation[6]. We introduced a dynamic measurement method, using sonoelasticity, which was previously validated on a plastic pelvis model[7] and on embalmed human pelvises[8]. In human pelvises a range of stiffnesses was created from immobilization with screws up to cutting the SI ligaments. With the help of this method a significant stiffness difference could be determined in SI joints. The results showed the potential of this method for in vivo application, because of its non-invasive character and the fact that the applied high frequency and low amplitude vibrations without resonance are not harmful for human health[9]. The next step in the development of this method is to answer the following questions; can the method be applied in vivo?; what is the reproducibility in vivo?; and what is the range of stiffness that occurs in healthy volunteers?

**Method and Materials**

Fourteen healthy volunteers were included in the study; anthropometric data are given in table 3.1. All subjects were females without any problems at low back and pelvic area for at least one year, and no regular use of alcohol or pain killers. Before the experiments the data were collected from the subjects by means of a questionnaire consisting of 22 questions. The questions related to life pattern (occupation, sports), health (low back and pelvic pain) and family details (number of pregnancies, children etc.). Informed consent was given by the subjects.

Table 3.1. Anthropometric data of the subjects. S smokers, P anti-conception pill user, C presence of children.

| Subject | Age (year) | Weight (kg) | Height (cm) | | S | P | C |
|---|---|---|---|---|---|---|---|
| P1 | 35 | 65 | 175 | | - | + | - |
| P2 | 31 | 68 | 180 | | - | + | - |
| P3 | 28 | 66 | 178 | | - | + | - |
| P4 | 29 | 65 | 172 | | - | - | - |
| P5 | 35 | 64 | 173 | | - | - | - |
| P6 | 31 | 70 | 173 | | - | + | - |
| P7 | 29 | 69 | 180 | | - | + | - |
| P8 | 28 | 65 | 186 | | - | + | - |
| P9 | 40 | 60 | 168 | | + | - | + |
| P10 | 28 | 68 | 172 | | - | + | - |
| P11 | 25 | 72 | 180 | | + | - | - |
| P12 | 28 | 59 | 176 | | + | + | - |
| P13 | 20 | 65 | 160 | | - | - | - |
| P14 | 33 | 65 | 173 | | - | - | + |
| avg | 30 | 65.8 | 175 | Total | 3 | 8 | 2 |

The experimental set-up was similar to that used in earlier studies on sonoelasticity measurements on SI joints (model[7] and cadavers[8]). A vibrator, Derritron VP3 with a frequency range between 1.4 and 20000 Hz and a Colour Doppler Imaging (CDI) apparatus (Philips AD1) was used in the experiments. Vibrations with 200 Hz of frequency were applied unilaterally to the anterior superior iliac spine of the healthy volunteer in supine position (Figure 3.1). The subjects were in a prone position with relaxed muscles, especially the large gluteus, biceps femoris and erector spine muscles. The vibrations propagate with a spheric distribution in the pelvis up to the SI joint area. The vibrations are picked up as moving reflectors by the CDI transducer which covers both sides of one SI joint (Figure 3.2). The intensity of the vibration pixels of the sacrum and ilium appear simultaneously on the monitor at high threshold values. The threshold indicates the necessary signal power to display in colour as motion. When the energy of the Doppler signal received back from the vibrating elements of the sacrum and ilium exceeds a certain level, these elements are displayed in red and blue on the CDI monitor. Below this level conventional grey-scale B-mode images are displayed. The height of the level can freely be set by the operator by means of the threshold button present on the control panel of the CDI apparatus. Using the threshold button allows to perform comparative vibrating energy measurements between the sacrum and ilium as follows. First a threshold level is found at which the Doppler colour image of the vibrating sacrum disappears and changes to grey scale. At the next step a second threshold level is found for the ilium. Since the threshold levels are directly related to the vibration energies of the bone, a large difference between the threshold levels of the sacrum and ilium indicates a large loss of energy through the SI joint, which means an instable joint. A small difference, or the absence of it, is an indirect indication of a stiff joint.

Measurement of the left and right SI joints were repeated three times for each side at each session. Between each measurement the subject had to stand up and perform some light exercises with

arms and legs. In total three different sessions took place on different days and at randomly chosen times. The data collected from each pelvis were divided into two groups: i.e. the left and right SI joints. One-way Analysis of Variance-test was applied separately to both groups to test the measurement reliability. The T-test for Paired Samples was applied to detect significant differences between the two groups. For all tests a significance level of 0.05 was chosen.

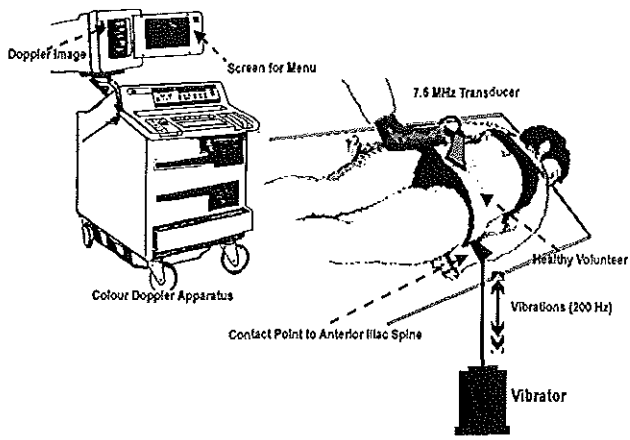


**Figure 3.1.** In-vivo experimental set-up.

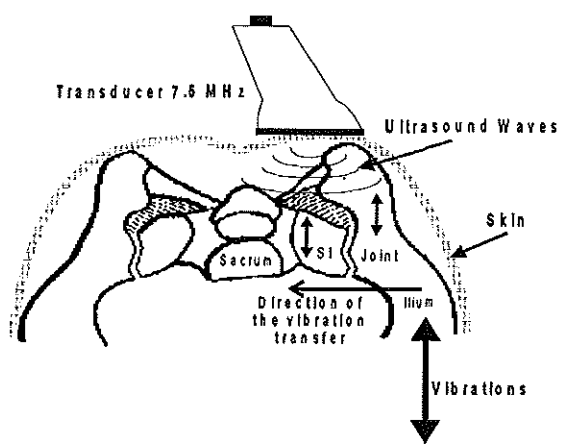


**Figure 3.2.** Positioning of the transducer on the sacroiliac joint area.

**Results**

From the questionnaire, it appeared that in the group of 14 healthy female subjects there were three smokers, eight used anti-conception pills, and two gave birth to one or more healthy children (Table 3.1). Table 3.2 presents the differences between the respective minimum thresholds which were recorded for all subjects, i.e. Pelvis 1 to 14. The threshold difference values representing SI joint stiffness range from

0 to 15. As an example: test person P1 had a low threshold difference (0) at both left and right SI joints, which is a measure for high stiffness of both SI joints. Test person P6 shows a large threshold difference indicating unstable SI joints. In test person P10 the right SI joint seems stiffer than the left one. One-way Analysis of Variance reveals that the reproducibility of the method is good because all left SI joints showed a Reliability Coefficient of 0.97 and all right SI joints 0.94 (97% and 94%, respectively). The variation in reproducibility for the left side was intra-individual 0.39 and inter-individual 11.85, and for the right side 0.45 and 6.83, respectively. Variance analysis showed that the method can significantly discriminate different inter-individual stiffness. The T-test applied to the results from the left and the right sides showed no statistical significant difference with a P-value of 0.44.

Table 3.2. The differences in the threshold values seen in the sacroiliac joints in threshold units. Rep repetitions, P1 to P14 subjects, L left, R right, S1 to S3 sessions, avg average, SD standard deviation.

| | | Subjects | | | | | | | | | | | | | | | | | | | | | | | | | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | | P1 | | P2 | | P3 | | P4 | | P5 | | P6 | | P7 | | P8 | | P9 | | P10 | | P11 | | P12 | | P13 | | P14 | |
| | Rep | L | R | L | R | L | R | L | R | L | R | L | R | L | R | L | R | L | R | L | R | L | R | L | R | L | R | L | R |
| | 1 | 0 | 0 | 4 | 3 | 3 | 3 | 4 | 8 | 2 | 0 | 12 | 8 | 1 | 4 | 6 | 4 | 2 | 2 | 9 | 4 | 5 | 5 | 0 | 2 | 2 | 1 | 3 | 4 |
| S1 | 2 | 0 | 1 | 4 | 3 | 4 | 3 | 4 | 9 | 1 | 0 | 13 | 11 | 0 | 3 | 6 | 4 | 2 | 1 | 8 | 4 | 4 | 4 | 1 | 3 | 1 | 0 | 2 | 3 |
| | 3 | 1 | 0 | 4 | 3 | 3 | 4 | 4 | 7 | 1 | 1 | 12 | 10 | 1 | 4 | 6 | 3 | 2 | 1 | 8 | 5 | 5 | 4 | 0 | 2 | 1 | 0 | 2 | 4 |
| | 4 | 1 | 1 | 4 | 3 | 3 | 3 | 5 | 7 | 2 | 0 | 15 | 10 | 0 | 3 | 7 | 3 | 3 | 2 | 7 | 4 | 4 | 4 | 0 | 2 | 1 | 0 | 3 | 3 |
| S2 | 5 | 1 | 0 | 4 | 3 | 3 | 4 | 5 | 5 | 1 | 0 | 13 | 11 | 0 | 3 | 7 | 3 | 3 | 2 | 8 | 4 | 5 | 4 | 0 | 3 | 1 | 1 | 3 | 3 |
| | 6 | 1 | 0 | 5 | 3 | 3 | 3 | 4 | 6 | 1 | 1 | 13 | 10 | 1 | 3 | 7 | 4 | 2 | 2 | 9 | 5 | 5 | 5 | 1 | 2 | 2 | 1 | 3 | 3 |
| | 7 | 0 | 1 | 5 | 4 | 4 | 3 | 5 | 6 | 2 | 0 | 13 | 11 | 0 | 2 | 7 | 3 | 2 | 2 | 7 | 5 | 5 | 3 | 1 | 2 | 1 | 1 | 2 | 4 |
| S3 | 8 | 2 | 1 | 5 | 4 | 5 | 3 | 4 | 7 | 2 | 1 | 13 | 10 | 1 | 2 | 7 | 3 | 2 | 3 | 7 | 4 | 4 | 4 | 1 | 3 | 1 | 0 | 2 | 4 |
| | 9 | 2 | 0 | 4 | 5 | 4 | 3 | 4 | 6 | 1 | 1 | 12 | 11 | 0 | 3 | 7 | 4 | 2 | 2 | 6 | 5 | 5 | 4 | 0 | 2 | 1 | 0 | 2 | 3 |
| | avg | *0.8* | *0.4* | *4.3* | *3.4* | *3.5* | *3.2* | *4.3* | *6.7* | *1.4* | *0.4* | *12.9* | *10.2* | *0.4* | *3* | *6.6* | *3.4* | *2.2* | *1.8* | *7.6* | *4.4* | *4.6* | *4.1* | *0.4* | *2.3* | *1.2* | *0.4* | *2.4* | *3.4* |
| | SD | *0.7* | *0.5* | *0.4* | *0.6* | *0.6* | *0.4* | *0.4* | *1.1* | *0.5* | *0.5* | *0.8* | *0.9* | *0.5* | *0.6* | *0.4* | *0.5* | *0.4* | *0.5* | *0.9* | *0.5* | *0.4* | *0.5* | *0.5* | *0.4* | *0.4* | *0.5* | *0.5* | *0.5* |

**Discussion**

The inclusion criteria for the healthy population in this study were selected with regard to future studies on the etiologic relation of low back and pelvic pain to the biomechanics of SI joints. Chronic low back and pelvic pain, or pain sensitivity decreasing addictions (e.g. regular alcohol consumption and pain killers) were designated as exclusion criteria. Since steroid-based hormones can cause laxity in the joints in females[10,11], the use of hormonal drugs such as anti-conception pills is assumed to be an effective factor in the pelvic stability level. Although we collected and presented information on the possible effective variables by mean of the questionnaires, we did not investigate statistically their relation with the SI joint stiffness measured. The number of the subjects included in the study was too small to allow conclusions from these factors. The present study was restricted to reproducibility, while future studies will focus on pathology and other possibly effective factors. The wide range of variability on the amount of stiffness in the healthy subjects shows that the etiology of the low back and pelvic problem is not only due to hyper- or hypo-mobility in the SI joints.

The existing methods for stiffness assessment of SI joints have the disadvantage of not being reproducible, and the results depend on the testing method and on the experience of the examiner. The method introduced in this study uses apparatuses which are accurate. In this study assessment of SI joint stiffness with sonoelasticity was tested for in vivo applicability, and for reproducibility and variability. It was possible to re-design the experimental set-up from earlier studies[8] for in vivo application which showed that this method is applicable on human beings. Application on 14 healthy female subjects revealed that in vivo reproducibility is satisfactory. Only a small percentage (left 3% and right 6%) of the measurements were not reliable reproducible. Reproducibility, however, seems to reduce with lower

SI joint stiffness, which may be due to tension of the muscles which may affect the SI joint stiffness[12]. It is expected that the measured range from 0 to 15 would increase with more test persons. The maximum threshold value on the CDI apparatus is 27. However, the accuracy decreases above the value of 20 due to an increase of sensitivity for external influences such as vibrations in the air.

Based on the promising results of the application of this method on a plastic pelvic model[7], human embalmed pelvises[8], and healthy subjects we assume that future application in the clinic is feasible. However, some limitations exist. First, the set-up comprises two expensive components, the vibrator and the CDI apparatus. The present CDI equipment has been especially designed for blood flow examination. A less expensive apparatus, especially designed to assess SI joint stiffness, could be developed for this type of measurements in the future. The assessment of two SI joints took 20 minutes (without repetitions) in the present study. It is expected that technical development of the in vivo application will lead to a reduction in time and costs. Second, the position of the subject is currently restricted to the supine posture in order to ensure good contact with the vibrator and the biomechanically released SI joints. Testing in other postures, like sitting and standing, may provide a better understanding of the influence of passive structures (ligaments) under load and of muscle activity on SI joint stiffness.

## Conclusions

- The dynamic examination method using sonoelasticity for SI stiffness is reproducible in vivo;
- Healthy subjects experience no inconvenience with this non-invasive measurement procedure;
- Clinical application of this new technique seems feasible, particularly because it is user-friendly and would be less expensive in the future.


## Acknowledgements

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# Chapter 4

This chapter is mainly based on the following publication

HM Buyruk, HJ Stam, CJ Snijders, A Vleeming, JS Laméris, WPJ Holland. The measurements of sacroiliac joint stiffness in peripartum pelvic pain patients with colour Doppler imaging. (Submitted to Spine) 1996

# Measurement of Sacroiliac Joint Stiffness in Peripartum Pelvic Pain Patients with Doppler Imaging of Vibrations (DIV)

## Chapter 4

**Abstract**
Peripartum pelvic pain (PPPP) is a condition related to the pathology of sacroiliac (SI) joints after the exclusion of other causes, such as lumbar disc hernia or spondylolisthesis. Although routine physical examination and radiologic investigations (X-rays, computerised tomography and magnetic resonance imaging) provide information about the pathology of the lumbar spine and the pelvis, these techniques do not contribute to the understanding of abnormal biomechanics. In previous studies we introduced a new technique to assess SI joint stiffness using Colour Doppler Imaging (CDI) and vibrations. This technique is called Doppler Imaging of Vibrations (DIV). It was first tested on a plastic model, embalmed human pelvises and healthy volunteers. In the present study the measurements were done on a group of PPPP patients (n= 56) and a healthy control group (n= 45). A vibrator, Derritron VP3 and a Colour Doppler Imaging apparatus (Philips Quantum AD1) were used. A protocol similar to that published earlier by our group was followed. The difference in SI joint stiffness between the patient group and the control group were statistically tested by means of Wilcoxon's two sample test, the chi-square test and Student's t-tests. Both patients and healthy controls displayed hypo- and as well as hyper mobility, but there was no significant difference between the groups. There was a significant difference between the groups with regard to the average difference of SI joint stiffness between left and right. Asymmetric stiffness of the SI joints seems to be more directly related to PPPP than the stiffness level of a single SI joint.

**Keywords**
Low Back Pain, Peripartum Pelvic Pain, Sacroiliac Joints, Stiffness, Doppler, Instability

### Introduction

Low back pain (LBP) is a very common complaint which portrays a disorder causing restrictions in daily life activities[13]. Idiopathic LBP is described as a painful condition without diagnosed pathology, such as lumbar disc hernias or spondylolisthesis. Earlier studies suggested that idiopathic LBP can be related to pelvic biomechanics and, especially, the stiffness of the sacroiliac (SI) joints[26,27]. Although routine physical examinations[12,14] (range of motion and other special tests) and laboratory investigations (X-rays, computerised tomography and magnetic resonance imaging)[20] provide information about the pathology of the lumbar spine and SI joints, these techniques do not contribute to the understanding of abnormal pelvic girdle biomechanics. Stiffness of the SI joints is one of the parameters involved in the biomechanics of the low back and the pelvis. There are different theories about the causes and consequences of hyper- and hypo-mobility in SI joints[1,6,10,15,16,17,24,25].

The change in the pelvic girdle stiffness during and after pregnancy has been described as a cause of peripartum pelvic pain (PPPP)[8,18,22,23]. The restoration of SI joint stiffness can be delayed temporarily or permanently after childbirth. Therefore, in vivo assessment and follow-up of the SI joint stiffness is important for clinical and research purposes. However, all the existing non-invasive clinical tests, such as the Gillet test, for the assessment of pelvic stiffness are subjective and unreliable[7,8,9,19,21]. This led us to the development of a new non-invasive technique using dynamic excitation which was first tested on a plastic model[5]. Subsequently, this technique was applied on cadavers and on healthy subjects[3,4]. The technique allows for the quantification of SI joint stiffness by comparing the intensity of the vibrations between iliac and sacral bones. The intensity of the vibrations is measured by means of Colour Doppler Imaging (CDI) and this technique is called

Doppler Imaging of Vibrations (DIV) by our group.

The results of the previous experiments on healthy subjects revealed that DIV is reliable and reproducible in the assessment of SI joint stiffness[4]. The research question of the present study was: are SI joint stiffness levels of PPPP patients different from those of healthy subjects. Therefore, we compared the left and the right SI joint stiffnesses of PPPP patients with those of a healthy control group.

**Materials and Methods**

A group of PPPP patients (n= 56; aged between 24 and 54 years) and a control group (n= 45; aged between 20 and 42 years) participated in the study. The patient group and control group had an average of 1.46±0.7 and 0.11±0.4 children, respectively (Figure 4.1). The inclusion criterium for the patient group was the existence of pelvic and/or low back pain during at least three months or more after delivery. The diagnosis was made by a physician who was not involved in this study. The clinical data were collected from the hospital files after the DIV measurements were performed. The control group consisted of females without a history of idiopathic pain in the region of the pelvis and lower back. Informed consent was given by all subjects.

**Table 4.1.** Age, number of children and measurement results from the patients and controls.

| | Patient (n=56) | | Control (n=45) | |
|---|---|---|---|---|
| | Mean | SD | Mean | SD |
| Age | 33.1 | 5.7 | 27.6 | 4.9 |
| Children | 1.46 | 0.7 | 0.11 | 0.4 |
| Left SI stiffness | 2.57 | 2.7 | 3 | 2.6 |
| Right SI stiffness | 2.8 | 2.9 | 2.8 | 2.5 |
| Absolute difference | 3.38 | 2.3 | 0.73 | 1.2 |

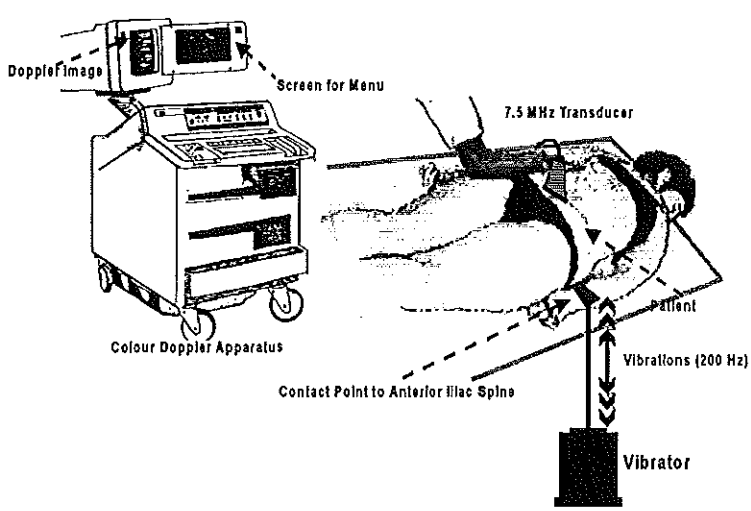


**Figure 4.1.** Experimental set-up.

The experimental set-up was similar to that used in earlier studies on SI joint stiffness[3,4]. A vibrator, Derritron VP3 and a CDI apparatus (Philips Quantum AD1) were used. Vibrations with a frequency of 200 Hz were applied unilaterally to the anterior superior iliac spine[5] (Figure 4.1). The subjects were asked to lay prone on an examination table with all the muscles relaxed. The vibrations propagate with a spheric distribution in the pelvis up to the SI joint area. The vibrations are picked up from ilium and sacrum by a CDI transducer which covers both sides of one SI joint (Figure 4.2). The intensity of the vibration pixels of the sacrum and ilium appears simultaneously on the monitor at high threshold values. The threshold indicates the necessary signal power to display in colour as motion. When the energy of the DIV signal received from the vibrating sacrum and ilium exceeds a certain level, they are displayed in red and blue on the CDI monitor. Below this level conventional grey-scale B-mode images are displayed. The height of the level can freely be set by the operator by means of the threshold button present on the control panel of the CDI apparatus. Using the threshold button allows to perform comparative measurements between the sacrum and ilium as follows. First, a threshold level is measured for sacrum at which the colour pixel of the vibrating sacrum disappears and changes to grey scale. Next, a second threshold level is measured for the ilium. Since the threshold levels are directly related to the vibration energies of the bone, a large difference between the threshold levels of the sacrum and ilium indicates a large loss of energy through the SI joint, which means an unstiff joint. A small or absent difference is an indication of a stiff joint.

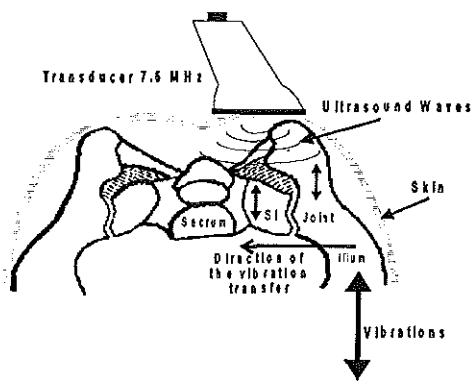


Figure 4.2. Application of the transducer on the sacroiliac joint.

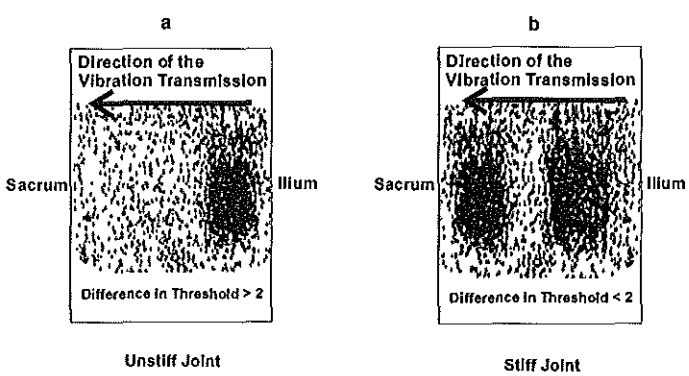


Figure 4.3. Typical DIV image of a stiff and an unstiff joint.

The patient group and the control group were compared in three ways:

**1.** To test the difference in the prevalence of only hyper mobility in SI joints between patients and healthy subjects, the mean joint stiffness of each group was compared using Wilcoxon's two sample test. This was done for the left and right SI joint separately, and in combination (mean of left and right joint).

**2.** To test the difference in the prevalence of hyper- or hypo-mobility between patients and controls, the results were grouped into the following categories: threshold difference; <3 hypo-mobility, 3 to 7 intermediate mobility, >7 hyper-mobility. Then the two groups were compared using the chi-square test on the 2x3 table. Again, this was done for the left and right joint separately, and in combination (mean of left and right joint).

**3.** To test the prevalence of the asymmetric SI joint stiffness, the absolute difference in mobility between the left and right SI joint was computed. Then these differences were compared between patients and controls using Wilcoxon's two sample test. The same test was repeated with the age stratification of 10 years.

In order to establish whether or not there is a relation between the location of the pain and the unstiff (threshold>7) SI joint, this was tested with the Student's t-test in the patients (n= 27) who described their pain as unilateral dorsal pelvic pain. For all tests the significance level was set at 0.05.

**Results**

A drawing of typical DIV image from a stiff and an unstiff joint is given in figure 4.3. In DIV images of the SI joints with threshold differences lower than 2 Doppler pixels appeared simultaneously on ilium and sacrum, while for larger differences pixels appeared first on the ilium and later on the sacrum. Table 4.1 presents the information on age, number of children and the results of DIV measurements of patients and of healthy controls. Mean (sd) age for patients and controls was 33.1 (5.7) and 27.6 (4.9) years, respectively. The difference in age and the number of children was statistically significant.

The statistical significance of the three comparisons were as follows:

**1.** In both groups low as well as high threshold levels were seen (Table 4.1). Figure 4.4 (right SI) and figure 4.5 (left SI) show the equal distribution of stiffness in the patient and the control group. No statistical significant differences in mean mobility were found between patients and controls (left SI: $p=0.78$, right SI: $p=0.99$, left and right SI averaged: $p=0.57$). The mean SI joint stiffnesses for patients and controls were: 2.7, 3.0, (left SI) and 2.8, 2.8 (right SI), respectively.

**2.** Furthermore, it appeared that the patients did not show significantly more hyper- or hypo-mobility than the healthy controls (left SI: $p=0.75$, right SI: $p=0.82$, left and right SI averaged: $p=0.27$).

**3.** However, a highly significant difference was found between the groups with regard to the absolute difference of SI joint stiffness between left and the right ($p<0.00001$), patients showing much larger asymmetry of stiffness than controls (means: 0.73 and 3.38, respectively; for distribution see figure 4.6 ). The same test with age stratification of decades also presented significant differences of $p<0.00001$.

There was no significant relation between the location of the pain and the unstiff side ($p=0.22$).

**Discussion**

This paper reports on the first application on patients of a new instrumented method (DIV) to assess SI joint mobility. The experimental set-up could be satisfactorily used in vivo, especially in patients with low back and pelvic pain.

The clinical application is facilitated by the fact that the positioning of the vibrator with respect to the ilium is not critical, because the vibrations are spherically distributed in the pelvic bones. Furthermore, the positioning and angulation of the CDI transducer is not critical because only the difference in vibration intensity between the left and right side of the SI joint is of interest.

In clinical practice pelvic instability is considered as a pathological condition. In the control group,

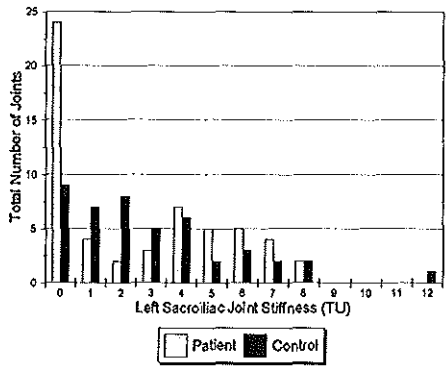


**Figure 4.4.** Stiffness levels measured in threshold units (TU) from L sacroiliac joints of control and patient groups.

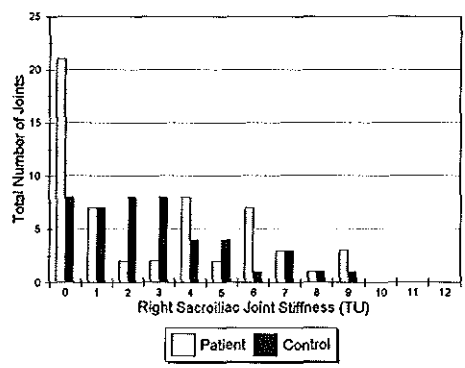


**Figure 4.5.** Stiffness levels measured in threshold units (TU) from R sacroiliac joints of control and patient groups.

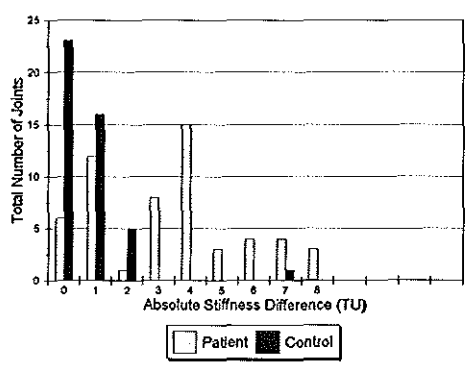


**Figure 4.6.** The distribution of absolute stiffness differences in threshold units (TU) between R and L sacroiliac joints.

however, unstiff joints as well as stiff joints are present. Therefore instability is not necessarily an indication of pathology and unstiff joints may be free of symptoms. The relation between the scientific term stiffness and the clinical term mobility was investigated in our pelvic model study[5]. It was demonstrated that a hyper-mobile SI joint was also mechanically unstiff and, vice versa, a hypo-mobile SI joint was stiff.

The main conclusion from this study is that in the patient group (independent of age) a symmetrical stiffness of the left and the right SI joint was formed. An unexpected result was that a range from hyper- to hypo-mobility of both SI joints was found in the healthy controls. This does not meet the conclusions reported in literature concerning SI joint problems. Some authors report hyper-mobility, ie instability of SI joints and symphysis pubis, as a major cause of low back pain[28]. However, hypo-mobility is also regarded as pathological according to some reports in literature[2,11,25]. The striking finding with regard to asymmetry in our patients, however, does not match with the reports in literature which point to asymmetry in pain symptoms[9,21]. The result of the present study contradicts with these authors who take the manually assessed mobility of one single SI joint as a diagnostic tool. Equal stiffness of the SI joints seems to be important for normal functioning. Some authors refer to the mechanism of damping and shock absorption[11]. They claim that asymmetrical shock absorption increases the risk of strain of the SI joints which causes discomfort or pain. However, shock absorption is required in walking, whereas most pain complaints are not reported during walking, but during standing and sitting[18]. Speculating about the consequences of asymmetric stiffness, the pathophysiology may be related to disturbed propriocepsis of the muscle groups that support SI joint stiffness; or may be the disturbed biomechanics of the lumber spine.

**Conclusions**

- A possible diagnostic tool is achieved that would contribute to the objective diagnosis of PPPP;
- DIV is easy to apply and non-invasive;
- In the healthy control group the left and right SI joint stiffness levels were generally symmetrical and in a range from loose to stiff;
- In the patient group a significant difference was found between left and right SI joint stiffness levels;
- Asymmetric stiffness of the SI joints seems to be more directly related to low back pain and pelvic pain than the stiffness level of a single SI joint.

**Acknowledgement**

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# Chapter 5

This chapter is mainly based on the following publication

HJ Stam, HM Buyruk, JS Laméris, CJ Snijders. Colour Doppler Ultrasound in Dynamic imaging of the Musculoskeletal System. Journal of Rehabilitation Sciences 1994 7(2): 49-52.

# Colour Doppler Ultrasound in Dynamic Imaging of the Musculoskeletal System: a preliminary report

## Chapter 5

**Abstract**
Most imaging techniques used to examine the musculoskeletal system, such as X-ray, CT scan and MRI produce mainly static images. Although Colour Doppler Imaging (CDI) is generally used in the examination of blood flow, it may offer the opportunity to study the dynamic characteristics of tendons and muscles. The purpose of this study was to explore the possibilities of the application of CDI in the imaging of the musculoskeletal system. A Philips Quantum Doppler was used to study the functional anatomy of the musculoskeletal systems of healthy subjects. This preliminary report on colour Doppler examination is focused on the imaging of wrist and hand. The results indicate that CDI offers a promising approach to image dynamically the tendons and muscles in the musculoskeletal system.

**Keywords**
Muscle, Tendon, Colour Doppler Imaging, Dynamic imaging

### Introduction

In rehabilitation medicine a thorough examination of the structures of the musculoskeletal system plays a pivotal role. Besides the physical examination, several imaging techniques are available for the physician in order to investigate the anatomy, the function and the pathology of bones, tendons, nerves, blood vessels and muscles[1].

The imaging modalities for the musculoskeletal system in general are: X-ray, CT scan, and MRI[2,3,4]. The advantages of these techniques are that they are available in almost all hospitals and that radiology departments have considerable experience in the application of these techniques. A significant disadvantage, however, is that since they are static techniques[5,6] they fail to give information about the functioning of the musculoskeletal system. X-rays display only bony structures and are possibly harmful for the patients. CT scans and especially MRI show bones and soft tissues but, so far, the patient has to be instructed not to move in order to improve the definition of the image.

EMG is a more suitable laboratory technique to examine the activity of the locomotor system. However, only electrical signals of the muscles and nerves become available and in the human body invasive needle EMG has to be used to determine the function of small muscles of the musculoskeletal system. The laboratory techniques for active examination of the locomotor system are restricted to EMG and some other measurements, especially for isolated muscle strength[7]. The analysis of the displacement direction and speed, and possible adhesions of separate tendons in vivo is still difficult or impossible with most of the current imaging and measurement techniques. Because of the restrictions of the existing techniques, we employed a new method to analyze the musculoskeletal system in more detail.

The combined use of sonography and the colour display of Doppler shifts from moving reflectors is one of the major developments in medical diagnostic procedures in recent years[8]. In clinical practice Colour-Doppler Imaging (CDI) is normally used for the examination of blood flow[9]. The physical base of CDI is the Doppler shift theory and, therefore, CDI has the ability to show the movement of structures other than red blood cells. CDI seems to be the appropriate technique to study the speed of displacement, the direction of movements, and the anatomy of tendons and muscles in the wrist and hand region. This issue, however, has not been previously addressed in the literature. Therefore, the aim of this study was to determine the applicability of CDI in the dynamic examination of the anatomical structures of the wrist and hand region.

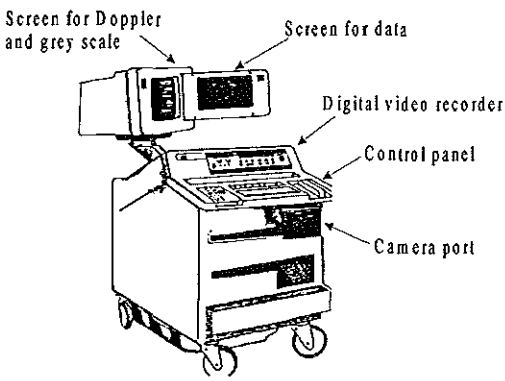


**Figure 5.1.** Colour Doppler Ultrasound (Quantum AD1) equipment.

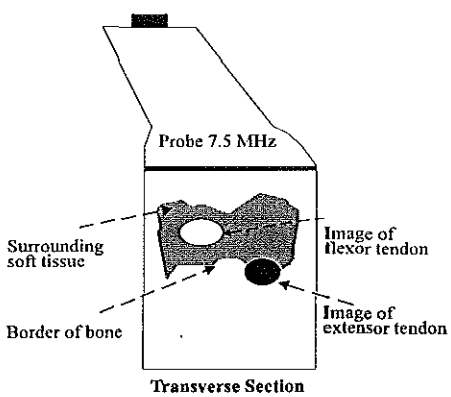


**Figure 5.2.** Diagram for positioning the probe parallel to the displacement.

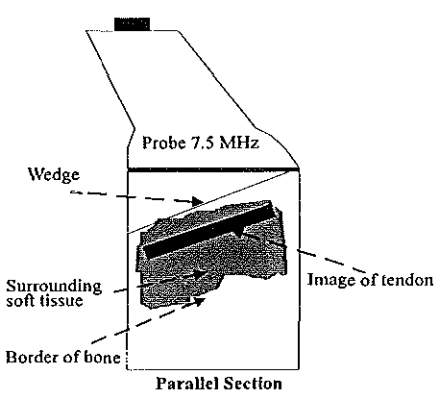


**Figure 5.3.** Diagram for positioning the probe vertical to the displacement.

**Materials and Methods**

The anatomy and function of the musculoskeletal system of the wrist and hand have been studied using a Philips Quantum Colour Doppler Ultrasound AD1 (AngioDynograph 1) with software version 2.5 (Figure 5.1). Five healthy volunteers ages between 25 and 42 were used as subjects for the experiments. A procedure similar to blood flow examinations was followed. Many of the anatomical structures of the wrist and hand are superficial, hence the "small parts" - examination option was chosen. The 7.5 MHz linear probe and contact gel were used. The apparatus worked on the "slow flow" - mode to discriminate low frequency shifts. With an adjustment button (threshold) it is possible to determine the best sensitivity for the apparatus to display slow movements. The structures were scanned parallel (Figure 5.2) and transversal (Figure 5.3) to the direction of their movement. Since no Doppler shifts can be measured if the angle between the soundbeam and the direction of the movements is 90 degrees, an 18 degree wedge between the probe and the skin was used when the muscles and tendons parallel to the skin were examined. The transversal and parallel images of a structure were taken along its tract on different levels. The dynamic images were recorded on video-tape for further study and analysis. In this paper we present, as examples of the CDI-technique, the results of images obtained from the m. flexor pollicis longus (long flexor of the first finger) and the m. flexor digitorum communis superficialis III (third tendon of the common superficial finger flexor). These structures were scanned parallel to their tract at the level of the wrist, the carpal tunnel, metacarpal bones and the finger during flexion and extension. Different measurement options for Doppler shift, such as spectrum analysis and velocity, that are available on the apparatus were used to check the reaction for displacements of the desired structures. In the framework of this preliminary report we will restrict our results and discussions to the images and the future of the technique.

**Results**

In all subjects the muscles and tendons in the wrist and hand region were scanned. The apparatus was sensitive to all of the muscle displacements. The quality of the images were similar in all of them; therefore, we present one example for each scanning plane. A parallel scan of the displacement of the m. flexor pollicis longus at the level of the I. metacarpal bone is shown in figure 5.4; the relevant structures are indicated on the figure. The sliding of the tendon was displayed as blue (the colour is dependent on the adjustment of the apparatus) during contraction and red during its relaxation or contraction of its extensor. The other non-moving anatomical structures are displayed in grey scale tones according to their echo-reflective properties.

A transversal section of the m. flexor digitorum communis III. at the level of the MCP joint during flexion of the III. finger is shown in figure 5.5. The tendon can be easily distinguished from the surrounding tissues and the apparatus displays red or blue according to its direction. At the MCP joint level it is also remarkable to observe the lateral bundle of the extensor tendon from the same finger in contrast colour during flexion in the same image. Although it is not presented in this image it is also possible to scan all the finger flexors in the same view with all fingers flexed at the same time.

Figure 5.6 shows the displacement of the thenar muscles which are scanned transversally to their contraction axis. In the images obtained from the muscles, the form of the muscle as grey scale is seen clearly and the Doppler shift during the contraction was superimposed. The images give a good impression of the direction of the displacements occurring under the probe.

**Discussion**

At present the major imaging techniques available for the evaluation of musculoskeletal system are static methods. Grey-scale real-time sonography permits to follow dynamic activity of the locomotor system; nevertheless, the amount and quality of the information is restricted. Electromyography is very helpful in determining the muscle activity, but non-invasive superficial applications are not sufficiently precise for small muscles. The development of a new technique to image the functions of

the locomotor system can fill the gap between static imaging techniques and electromyography.

Analysis of the locomotion of certain parts of the body seems to be possible with Doppler sonography. Colour Doppler instruments are designed to study blood flow. In these instruments the two-dimensional Doppler shift or time shift information is colour coded and superimposed on the two-dimensional grey-scale real-time anatomic display. Several instruments use image processing algorithms that suppress Doppler shifts at pixels where strong echoes are detected on the grey-scale mode. Tendons and muscles are, however, echo-poor, especially when the angle between the soundbeam and the fibres of these structures is less than 90 degrees. Since angles of less than 60 degrees are also needed for optimal Doppler shift detection, these image processing algorithms do not hamper the colour imaging of these structures. The imaging results are satisfactory, especially for superficial structures. However, there are some limitations in detecting slow movement in deeper areas. Therefore, our trials concentrated on examination of superficial regions such as wrist, hand, ankle and foot. By correcting the vertical positioning of the probe to the direction of the movement with a stand-off wedge we were able to compare the function of tendons or muscles in regions such as wrist and carpal tunnel. The size of the probe and the poor definition of the colour display was an impediment to properly imagining the fine functional anatomy of wrist and hand region.

The colour flow display does not provide all the details of the Doppler spectrum that spectral display does; it only presents the direction of movement and the mean velocity. However, after angle correction, frequency shifts from each part (sample volume) of the tendon or muscle can be displayed as movements in cm/sec by using spectral analysis with pulsed Doppler (PD)[10].

Spectrum analysis and PD velocity calculations gave us an idea about the effectiveness of the locomotion, which can also be used as a criteria for the muscle functioning in the future. Examinations of superficial muscle areas provided the discrimination of the isotonic muscle activity. The grey scale was helpful in determining the muscle borders and the colour over these images determined the function.

In the future, this technique will assist in diagnosing and locating tendon pathologies like tendon rupture, adhesions, lumps, trigger finger, etc.. It might also be possible to get valuable information concerning the muscle pathologies, such as inactivity, rupture, adhesions and haematomas, which will accelerate the diagnosis time. In our opinion this examination technique is useful for research as well as for clinical departments such as surgery and rehabilitation. The technique can be easily performed in every radiodiagnostic department which possesses a combined colour Doppler ultrasound.

## Conclusion

Although we conclude that the application of CDI in the locomotor system is satisfactory and promising we consider further development of the CDI necessary to optimise the images and quantitative aspects of moving structures of the locomotor system. At present the software and probes of CDI are especially developed for blood flow examinations and the sensitivity is completely designed to the flow of the blood. However, the definition of colour Doppler display could be improved and especially designed probes and developed software filters would contribute considerably to the value of the technique. We expect that in the near future the software can be specially adapted to the range of velocities of the tendons and muscles which we are interested in.

## Acknowledgements

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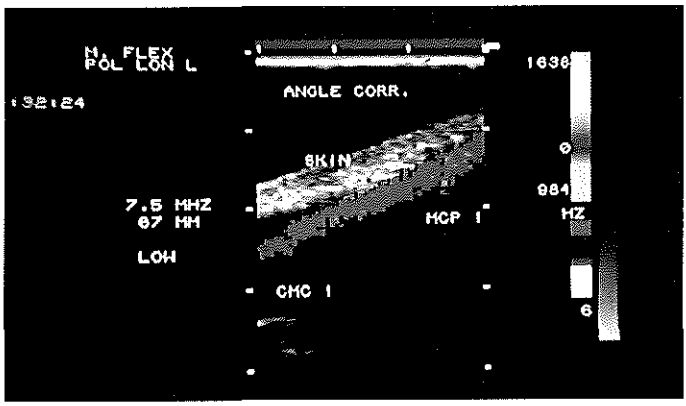


**Figure 5.4.** Image obtained from the contraction of the m. flexor pollicis longus at the level of the I. metacarpal bone with the probe positioned parallel to the contraction.

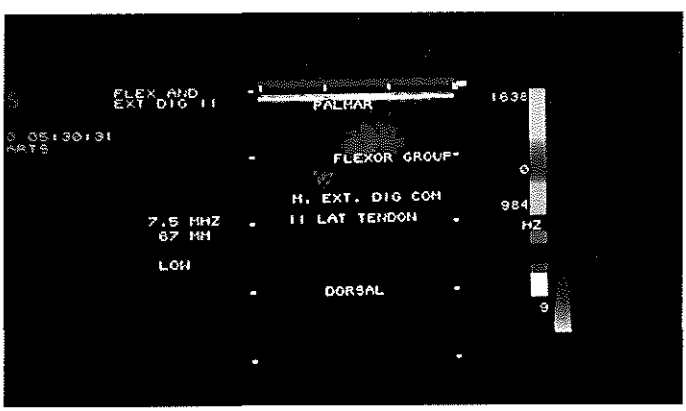


**Figure 5.5.** Image obtained from the contraction of the m. flexor digitorum superficialis III. tendon at the level of the MCP joints with the probe positioned vertical to the contraction.

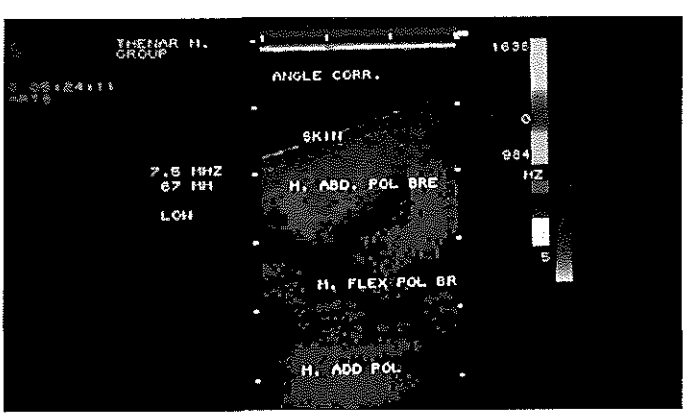


**Figure 5.6.** Image obtained from the contraction of the thenar muscle group at the level of the mid- metacarpal bones with the probe positioned vertical to the contraction.

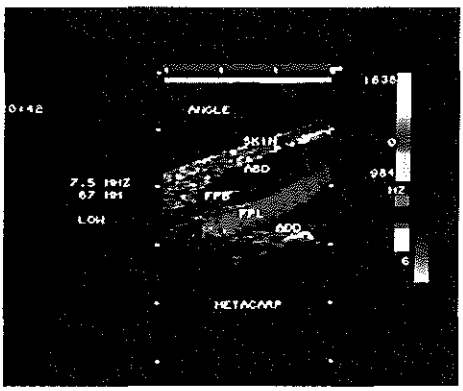


Figure 6.3. Parallel section; normal anatomic image of the FPL tendon at the metacarpal level in the patient with CMC I joint arthrosis.

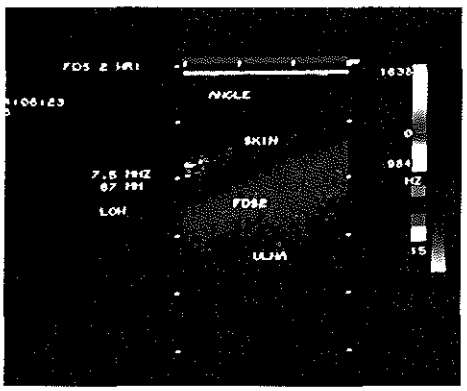


Figure 6.4. Parallel section; displacement of the FDS II tendon at the wrist level six months after the brachial plexus trauma.

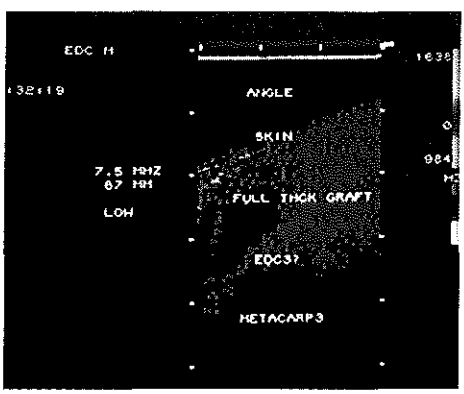


Figure 6.5. Parallel section; adhesion of the EDC III tendon to the forearm flap at the metacarpal level.

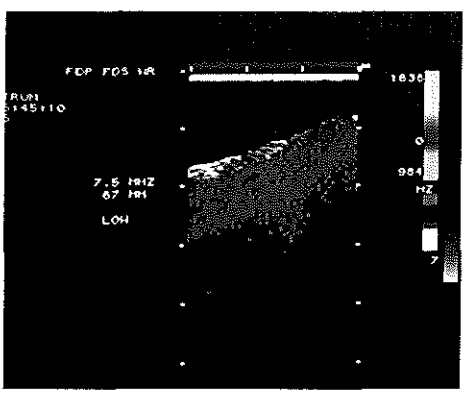


Figure 6.6. Parallel section; adhesion of FDS III and FDP III in the scar tissue after a trauma at wrist level.

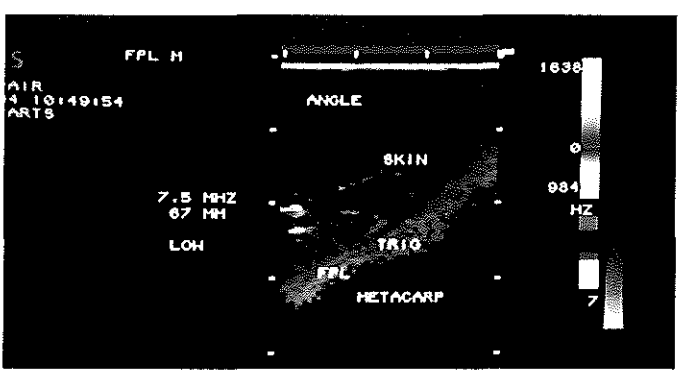


Figure 6.7. Parallel section; displacement of the FPL through the A1 pulley in the patient with trigger finger.

# Chapter 6

This chapter is mainly based on the following publication

HM Buyruk, HJ Stam, JS Laméris, HA Schut, CJ Snijders. Colour Doppler Ultrasound Examination of Hand Tendon Pathologies J. Hand Surgery 21B: 4: 469-473.

# Colour Doppler Ultrasound Examination of Hand Tendon Pathologies: a preliminary report

## Chapter 6

**Abstract**
Apart from real-time sonography, in most of the imaging techniques, used to examine the musculoskeletal system, such as X-ray, computerised tomography (CT) scan and magnetic resonance imaging (MRI) produce mainly static images. Although Colour Doppler Imaging (CDI) is generally used in the examination of blood flow, it may offer the opportunity to study the dynamic characteristics of tendons and muscles. The purpose of this study was to examine the unilateral flexor and extensor tendon pathologies of the hand and compare these images with the healthy side. A Philips Quantum Doppler was used to study patients with tendon adhesions and trigger finger. The normal and pathological tendons were scanned at three anatomical levels; wrist, metacarpal and proximal phalanx. The results obtained from the comparison of pathological images with normal ones indicate that CDI offers a promising approach to image dynamically and, in the future, to quantify the function of the tendons and muscles in the musculoskeletal system and to define their pathologies.

### Introduction

In the clinical practice of hand surgery and hand therapy, a thorough examination of relevant structures of the musculoskeletal system is essential. Apart from physical examination, several imaging techniques are available to investigate the anatomy, function and pathology of bones, tendons, nerves, blood vessels and muscles.

Imaging modalities often used for the musculoskeletal system are: X-ray, ultrasonography, computerised tomography (CT) scan, and magnetic resonance imaging (MRI; PUTMAN and RAVIN, 1988; KUHNS et al, 1989; SUTTON, 1993). A disadvantage is that they are all static techniques (RESNICK and NIWAYAMA, 1988; MURRAY, 1990), and cannot provide direct information about the function of the musculoskeletal system.

In superficial or invasive needle electromyography (EMG) only electrical signals of the muscles and nerves are obtained (KOTTKE and LEHMANN, 1990). Apart from EMG, dynamometry is an alternative for assessment of the musculoskeletal system. Neither EMG, dynamometry nor the above-mentioned imaging modalities give quantitative information about in vivo displacement speed, excursion and possible adhesions of tendons.

In clinical practice Colour Doppler Imaging (CDI) is normally used for the examination of blood flow (HAGEN-ANSERT, 1989; KREMKAU, 1993). The Doppler shift theory forms the physical base of CDI, giving it the ability to show movement of structures other than red blood cells. CDI appears to be the appropriate technique to study the speed and amount of displacement, and the direction of movements of the tendons and muscles in the wrist and hand region. This issue has recently been addressed in the literature (STAM et al, 1993; STAM et al, 1994). In that study, we used CDI as a new non-invasive method of imaging normal hand flexor tendon displacement. The tendons and muscles were imaged parallel and transverse to their displacements during contraction; the images showed that a clear definition of normal function is possible. The purpose of the present study is to determine the applicability of CDI on the normal and the pathological conditions of the flexor and extensor tendons.

### Materials and Methods

The function of the flexor tendons was studied in five patients with unilateral hand pathology

(Table 6.1). A Philips Quantum Colour Doppler Ultrasound AD1 (AngioDynograph 1) with software version 2.5 was used. Arthrodesis, open wounds and non-moving tendon adhesions were excluding criteria for the experiment. We present five examples of different pathological conditions: first carpometacarpal (1st CM) joint arthrosis; brachial plexus trauma; adhesion of forearm flap to third common digital extensor (EDC 3) tendon; increased resistance of deep (FDP 3) and superficial (FDS 3) digital flexor in the wrist after static treatment; and trigger thumb.

The patients were examined by CDI with a procedure similar to that applied to healthy subjects (STAM et al, 1994), and to superficial blood flow examinations. Patients were asked to flex and extend their fingers continuously. The "small parts"-examination option, 7.5 MHz linear probe and contact gel were used. In order to discriminate low frequency shifts, the apparatus worked on the "slow flow"-mode. The best sensitivity for the apparatus to display slow movements was determined with the threshold button. The tendons were scanned bilaterally parallel (Figure 6.1) at three different anatomical levels, and transversely (Figure 6.2) at the MP joint level, to the direction of movement. The three anatomical levels for parallel scans were wrist (proximal to radiocarpal joint), metacarpal and proximal phalanx. An $18^0$ stand-off wedge between the probe and the skin was used for parallel scanning. The images were recorded on video-tape for further study and analysis. The images of the pathological tendons were compared with the images taken from the opposite side of the same patient, and with the corresponding images from earlier studies with healthy subjects (STAM et al, 1994).

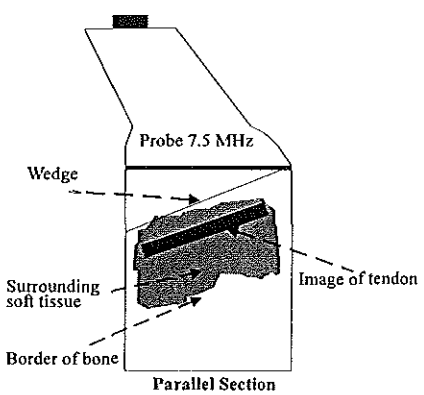


**Figure 6.1.** Drawing of a normal anatomic Doppler image of a tendon when the probe is positioned parallel to the displacement.

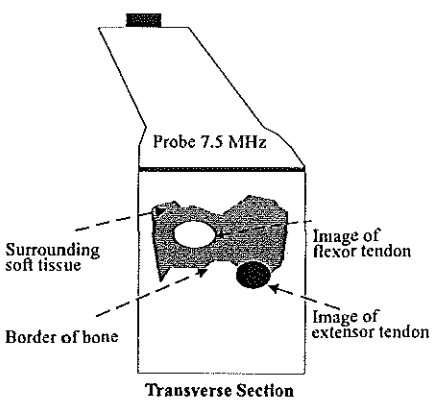


**Figure 6.2.** Drawing of a normal anatomic Doppler image of a tendon when the probe is positioned vertical to the displacement.

## Results

The apparatus was sensitive to all of the muscle and tendon displacements. Sliding of the tendon appeared as blue during contraction, and red during relaxation or contraction of the extensor. Other non-moving anatomical structures appeared in grey scale tones. The information obtained from the images was similar in all recordings, so we present the typical examples of pathology. There was no noticeable pathology in the images from transverse scanning.

Case 1 This patient had painful restricted mobility of first CM joint area. There was smooth sliding of the tendon of FPL (Figure 6.3). There was no evidence in the colour display of tissue traction in any of the scanned levels (wrist, metacarpal and proximal phalanx). These images were identical to healthy image examples.

Table 6.1. List of patients involved in the study.

| Patient | Sex Age (yrs) | Tendon | Anatomic Level | Pathology |
|---|---|---|---|---|
| 1 | F 65 | FPL | metacarphal | 1st CM arthrosis |
| 2 | M 25 | FDS2 | wrist | re-innervation |
| 3 | M 16 | EDC3 | metacarphal | adhesion |
| 4 | M 62 | FDS3, FDP3 | wrist | increased resistance |
| 5 | F 61 | FPL | proximal phalanx | trigger finger |

Case 2 A patient with brachial plexus trauma did not agree to the application of invasive needle EMG. Muscle activity of the index FDS was detected by CDI 3 months after the injury. The CDI displayed normal tendon displacement after 6 months, indicating recovery of muscle activity (Figure 6.4).

Case 3 A patient with dorsal hand trauma with a soft tissue defect developed an adhesion of the extensors to the forearm flap at the metacarpal level. CDI showed extensive tissue displacement (colour extensively superimposed on the grey scale image) under the probe from skin level to bone (Figure 6.5). There were no clear borders of tendon displacement.

Case 4 This patient developed abnormal resistance around the middle finger FDS and FDP tendon after repair. The connected displacement of both tendons appeared in CDI as a thick strip with irregular borders (Figure 6.6).

Case 5 A patient with trigger thumb, displayed thickening of the FPL image around the IP joint. In the CDI image it is clear that during contraction obvious traction appeared as a short displacement of the surrounding tissue, when the thickened tendon was passing the narrow A1 pulley (Figure 6.7).

**Discussion**

Case 1 is an example of a normal tendon image. Limited range of motion caused by the joint, or insufficient tendon length, are often misdiagnosed as tendon adhesion. The regular borders seen in CDI are a result of the normal sliding of the tendon in the tendon sheath.

Case 2 showed muscle recovery during the 6 months after the injury. The recovery of index finger FDS displacement could be monitored by CDI. The form of the tendon was similar to that of the normal side but the displacement speed was much slower.

Case 3 developed progressive adhesion of the forearm flap to all extensor tendons. Normally the CDI images of the extensors at the metacarpal level are highly disturbed by the contractions of the intrinsic muscles. However, it is easy to distinguish the regular thin strip of the undamaged extensor. There was no apparent tendon form.

Case 4 was treated incorrectly with a static splint for 6 weeks during the recovery period. Restriction of motion developed in the middle finger. The tendons often showed a thicker and irregular form than is normal at the wrist level CDI in patients with tendon lesions after static treatment. It is assumed that this situation would be a pro-adhesion or an increased resistance.

Case 5 was a typical example of irregular tendon displacement caused by trigger finger, ganglion, calcification or lumps. The simultaneous movements of the surrounding tissues were clearly visible as contrast colour, and the thickened line of the FPL tendon proved useful in determining the site of pathology.

Other than grey-scale real-time sonography, the major imaging techniques that exist for the evaluation of musculoskeletal system are static methods. Functional diagnostics as electromyography and dynamometry are not precise enough to quantify the activity of the fine muscles and tendons in the hand and wrist region. A new technique of imaging the functions of the locomotor system can serve as a useful addition to static imaging techniques and to electromyography. Since the demonstration of pathology is dynamic in the images it is not possible to display that feature with static pictures in this article.

Analysis of the movement of certain parts of the body is possible with CDI. In these instruments the two-dimensional Doppler shift or time shift information is colour coded and superimposed on the two-dimensional grey-scale real-time anatomical display. Several instruments use image processing algorithms that suppress Doppler shift at pixels where strong echoes are detected on the grey-scale mode. Tendons and muscles are echo-poor, especially when the angle between the sound beam and the fibres of these structures is less than $90^0$. Since angles of less than 60 are also needed for optimal Doppler shift detection, these image processing algorithms do not hamper the colour imaging of these structures. There are some limitations in detection of slow movement in deeper areas. Therefore, our trials focused on examination of superficial regions such as the wrist and hand. By correcting the vertical positioning of the probe to the direction of the movement with a stand-off wedge we were able to compare the function of the tendons or the muscles. The size of the probe and the poor definition of the grey-scale and colour Doppler display were an impediments to proper imaging of the fine functional anatomy of the wrist and hand region.

In the framework of this report we restricted our results and discussions to the pathological differences of the muscle function in the CDI images, and the future of this technique for the diagnosis of flexor and extensor pathologies in the hand. The pilot quantitative measurements, such as spectrum analysis, point velocity and point displacement distance similar to blood flow examinations, provided ideas regarding the effectiveness of CDI in locomotion.

From the results of this study it is clear that this type of examination can provide additional information for diagnosis, and even contribute considerably to the planning of treatment. The comparison of the images from the pathological extremity with those of the healthy opposite extremity and with other healthy subjects double-checked our conclusions over abnormality. Transverse imaging was done successfully only at the MP level because of the flat anatomy of the palmar surface which does not permit full contact with the probe at less than $60^0$.

In the future, we hope this technique will assist in diagnosing and monitoring tendon pathologies like rupture, adhesions, swellings and triggering. It is likely that valuable information can be obtained concerning the muscle pathologies, such as inactivity, rupture, adhesions and haematomas, which will shorten the diagnosis time.

Although we conclude that the application of CDI flexor tendon pathology in the hand is satisfactory and promising we consider further development of the CDI necessary to optimize the images and the quantitative aspects of moving structures of the locomotor system. Currently, the software, probes and sensitivity of CDI has been adjusted for blood flow examinations. The definition, probes, hardware and software of CDI should be improved for a better examination of the musculoskeletal system.

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# Chapter 7

This chapter is mainly based on the following publication

BS Cigali, HM Buyruk, CJ Snijders, JS Laméris, WPJ Holland, R Mesut, HJ Stam. In vivo flexor pollicis longus tendon contraction speed measurements with colour Doppler imaging (European Journal of Radiology in press) 1996.

# Measurement of Tendon Excursion Velocity with Colour Doppler Imaging: a preliminary study on FPL muscle

## Chapter 7

**Abstract**

**Purpose-**To study the use of Colour -Doppler Imaging (CDI) for the measurement of maximum and mean tendon velocity. Recent studies showed that CDI, normally used for blood flow examinations can be used for the imaging of tendons at the hand and wrist region. Although other modalities are available for imaging of the musculo-skeletal system, in vivo measurements of the velocity of tendon excursion are not possible.

**Method-** The flexor pollicis longus (FPL) tendon of 16 healthy volunteers was measured bilaterally at two levels (wrist and thenar). A splint from the fingers along the proximal lower arm was applied. The thumb was fixed to the splint from the first phalanx to allow flexion of the interphalangeal (IP) joint only. Pulsed CDI was used for the measurements. The maximum and mean velocities of the FPL tendon were measured at spectrum display mode during continuous voluntary contractions. At least ten sequential Doppler peaks (cm/sec) were recorded at every trial. The measurements were repeated three times. Paired t-test and correlation coefficients were calculated between levels on the same side and the opposite side.

**Results-** No significant differences were found between two levels of the same hand and of the opposite hand. As expected, the data revealed variations in the inter-individual tendon velocities.

**Conclusion-** The velocity of the excursion of the FPL tendon can be measured with CDI with good reproducibility. It is expected that velocity measurements can be used in the future for the assessment of other tendons affected by various disorders.

**Keywords**

Hand, tendon speed measurements; Ultrasound (US), Doppler studies

### Introduction

Maximum tendon excursion velocity depends on the condition of the tendon, the tendon sheath and the muscle belly and its innervation. Intrinsic and extrinsic muscle diseases and aging[1] cause a significant decrease in the maximal velocity of isotonic muscle contraction. Dynamometers can determine this phenomenon indirectly by measuring joint rotation velocity and exerted moments of force. This gives optimal results for single joint motions driven by one muscle, as in knee extension. In case of multiple joints and/or muscles it is more complex to measure the displacement velocity of a specific tendon. Specific maximum velocity measurements for tendons such as hand flexors, may be valuable for diagnosis, treatment and follow-up.

The power (P) output of a muscle is the product of muscle (tendon) force (F) and tendon velocity (v):

$$P=F.v\ (Nm.s^{-1})$$

The acceleration (a) of a tendon excursion depends on the mass (m) that must be propulsed by the given muscle force (F):

$$a = F/m\ (m.s^{-2})$$

The non-invasive accurate measurement of in vivo muscle and tendon velocity or the amount of excursion is not possible with the available techniques. The musculo-skeletal system can be imaged by using different techniques[2,3], including radiography, ultrasound[4,5], CT scan[6,7], and MRI. However, these techniques have the disadvantage that they fail to give quantitative information about the

movements of muscles and tendons. Although the non-invasive examinations with superficial EMG[8] and muscle dynamometry[9,10] are closely related to the function, it is hard to isolate individual muscle function. With grey scale ultrasound such functions can be visualised, but not quantified.

For the diagnosis of cardiac and vascular disorders sonography and the colour display of Doppler shifts from moving reflectors are often used[11]. Similar to such blood flow examinations we demonstrated that Colour-Doppler Imaging (CDI) can visualise muscle and tendon movements[12] and differences due to pathology[13]. The purpose of the present study is to show whether the non-invasive quantification of tendon velocity is possible with CDI and to test the reproducibility of tendon velocity measurements which will be executed on healthy volunteers.

**Materials and Methods**

Maximum and mean velocity was measured with a CDI on the flexor pollicis longus (FPL) tendon. Sixteen healthy right-handed volunteers (aged between 24 and 44, mean 33.3 years) participated in the study. A dorsal lower arm splint, made of Aquaplast, from fingertips to elbow was applied in order to immobilise the lower arm. The thumb was fixed, by a Velcro hook and loop attachment band, to the splint from the first phalanx to only allow flexion of the interphalangeal (IP) joint. A Philips AD1 (Angio Dynagraph 1) with software version 2.5 pulsed multi-channel CDI scanner was used. A procedure similar to blood flow examinations was followed. A 7.5 MHz linear transducer was used. An adjustment button (threshold) allowed to determine the best sensitivity for the apparatus to display slow movements. The location of the FPL muscle was determined with the help of the threshold button. An 18° wedge between the probe and the skin was used.

The measurement procedure was standardized as follows. The excursion of the FPL tendon was measured at two levels; the thenar and the wrist (distal radius). The exact measurement points were marked on the skin. Volunteers were asked to flex and extend their thumbs continuously with maximal speed. After a clear Doppler image of the excursion has been obtained in the threshold mode, the apparatus was switched to the spectrum mode (Figure 7.1). The angle between the ultrasound and the tendon was corrected with the angle correction option placed on the apparatus. The maximum and mean of the point velocity options were chosen from the spectral menu on the screen. When the maximum velocity selected the system overlays the maximum Doppler shifted frequency on the spectral waveform. The maximum velocity is coloured green so as to be differentiated from the spectral waveform and is a scrolling waveform along with the spectra. The maximum Doppler shift is defined as the frequency below which 95% of the frequencies fall. The apparatus then calculates the velocity values from the equation of Doppler shift and displays it on the screen. It can also calculate the mean velocity at the same time. The mean Doppler shift is defined as the centroid frequency. The maximum and mean velocities of the FPL tendon were measured in the spectrum display mode during continuous active movement. At least ten sequential Doppler peaks (cm/sec) were recorded at every measurement. The measurements were repeated three times bilaterally.

The measurement results were processed with a spreadsheet and a statistics program. The Coefficient of variance (CV) of the repeated measurements for each level was calculated separately for each subject. The CV of the measurements from all of the subjects were calculated from the average of all the measurements for each level. Paired T-test and correlation were used to demonstrate if there were any significant differences between the measurements of the levels and the sides. All statistical functions were applied to both measurement results of maximum and mean Doppler velocities. For statistical significance a level of $P<0.05$ was chosen.

**Results**

It was possible to measure the FPL maximum and mean displacement velocity in all subjects at all measurement sites with CDI. Doppler waves appeared as shown in figure 7.1. Table 7.1 presents the typical measurement results of one subject. Subject 2 displayed the highest averages of maximum velocities 17.77+/-1.36 cm/sec, 17.37+/-2.20 cm/sec, 15.43+/-0.95 cm/sec from the left hand thenar level (LHTL), left hand wrist level (LHWL), and right hand thenar level (RHTL), respectively. Subject 16

displayed the highest velocity from the right hand wrist level (RHWL) 14.80+/-1.22 cm/sec. The lowest velocities were recorded from LHTL 9.22+/-0.64 cm/sec of subject 13, from LHWL and RHTL 10.6+/-0.23 cm/sec and 8.41+/- 0.72 cm/sec, respectively, from subject 4, from RHWL 9.76+/-0.35 cm/sec of subject 5. CV of maximum and mean velocity measurements appeared to be high (low accuracy) for ten sequential wave peaks, but was low (high accuracy) for measurement repetitions (Table 7.2). The intra-individual differences between thenar and wrist and between right and left were statistically significant. The correlation coefficients were high and ranged between 0.60 and 0.82 (Table 7.3).

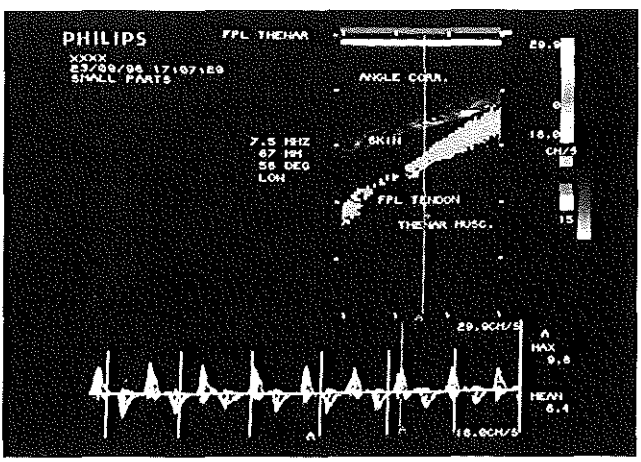


Figure 7.1. A typical CDI image of FPL tendon right hand wrist level.

**Discussion**

The FPL was selected because the tendon is superficial and therefore easy to image with CDI. Furthermore, it is the only flexor of the distal phalanx of the thumb and there are in general no inter-tendinous connections[14]. Some exceptions have been reported, such as anomalous tendon slips from the FPL to the deep flexor finger muscle which restricts FPL excursion[15].

FPL tendon maximum and mean velocities could be measured in all subjects. During fast thumb flexion/extension the maximum velocity was different for each repetition. We assume that these differences were caused by disoriented movements of the distal phalanx of the thumb. The frequency and the velocity of this movement do not only depend on muscular functions but also on neurological control during continuous voluntary movements. The assessment of the minimum number of repetitions necessary for the determination of a representative average is important for the design of future experiments. It is concluded that ten repetitions are sufficient because CV is low, indicating good reproducibility. The correlation between the velocity measurements at thenar and wrist was high which indicates that different measurement sites with different tendon section, tissue thickness, and density do not affect the results. Measurements of the left and the right FPL were correlated less, which can be attributed to factors as central or peripheral neural control. From the results it can be concluded that if a series of ten waves is recorded one time at one site, the maximal tendon excursion velocity is measured with high accuracy.

It is more difficult to obtain clear wave forms from the wrist area since the FPL muscle belly begins at the proximal carpal level and in the image it is displayed as a larger stripe in the CDI image. However, we observed no differences between the measurements of the maximum velocity at different sites of the same hand.

Although the maximum FPL velocity was slightly higher on the dominant side, the difference

Bilateral Maximum and Mean FPL Tendon Velocity Measurements of Subject 1 (cm\sec)

| P | Right Hand | | | | | | | | | | | | Left Hand | | | | | | | | | | | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | Thenar | | | | | | Wrist | | | | | | Thenar | | | | | | Wrist | | | | | |
| | Maximum | | | Mean | | | Maximum | | | Mean | | | Maximum | | | Mean | | | Maximum | | | Mean | | |
| | R1 | R2 | R3 | R1 | R2 | R3 | R1 | R2 | R3 | R1 | R2 | R3 | R1 | R2 | R3 | R1 | R2 | R3 | R1 | R2 | R3 | R1 | R2 | R3 |
| 1 | 9.9 | 11.7 | 11.3 | 5.4 | 5.9 | 6.3 | 11.3 | 9 | 8.1 | 5.9 | 5 | 4.5 | 11.7 | 12.3 | 11.9 | 7.7 | 7.8 | 7.6 | 11.3 | 11.5 | 11.5 | 6.3 | 6.1 | 6.5 |
| 2 | 14.9 | 10.8 | 10.9 | 7.2 | 5.9 | 5.9 | 10.8 | 10.4 | 9.9 | 8.6 | 5.9 | 5.4 | 12.2 | 14.1 | 12.8 | 8.1 | 8.3 | 7.9 | 10.8 | 11 | 11 | 5.4 | 5.5 | 5.6 |
| 3 | 9.5 | 10.8 | 9.5 | 5.4 | 5.9 | 5.4 | 12.2 | 10.4 | 9.9 | 5.9 | 5.9 | 5.4 | 13.1 | 13.5 | 13.5 | 7.2 | 7.4 | 8.1 | 10.4 | 10.9 | 10.9 | 5.4 | 5.4 | 5.8 |
| 4 | 14 | 12.8 | 11.6 | 6.3 | 6.8 | 6.5 | 9.5 | 9.9 | 9.5 | 5.4 | 5.4 | 5 | 15.8 | 13.6 | 15.2 | 9.9 | 8.2 | 9.4 | 13.1 | 13 | 13.5 | 6.3 | 6.2 | 6.5 |
| 5 | 12.6 | 10.8 | 7.2 | 6.8 | 5.9 | 3.6 | 14 | 8.6 | 9.9 | 5.9 | 5 | 5.4 | 12.2 | 15.1 | 11.9 | 7.2 | 8.5 | 6.9 | 10.8 | 10.9 | 10.5 | 6.3 | 6 | 5.7 |
| 6 | 8.1 | 12.2 | 12.2 | 4.5 | 5.9 | 6.8 | 9 | 9.9 | 10.1 | 5 | 5.4 | 5.7 | 11.3 | 12.6 | 11.9 | 7.7 | 7.8 | 7 | 12.2 | 12.9 | 13.4 | 6.3 | 6.3 | 6.8 |
| 7 | 15.3 | 10 | 12.4 | 7.7 | 5 | 6.9 | 10.4 | 9.9 | 10.3 | 5.9 | 5.9 | 5.7 | 11.3 | 13.1 | 12.6 | 7.2 | 7.6 | 7.2 | 10.8 | 11 | 11.4 | 5.9 | 5.8 | 6 |
| 8 | 14 | 11 | 12.6 | 6.8 | 5.9 | 7.2 | 11.7 | 9.5 | 10.4 | 5 | 5 | 6 | 10.4 | 12.9 | 12.5 | 6.3 | 6.9 | 7 | 11.7 | 12.4 | 11.7 | 6.3 | 6.1 | 6.1 |
| 9 | 11.3 | 11.7 | 12.8 | 5.9 | 6.8 | 7.4 | 9 | 11.3 | 8.1 | 4.1 | 5.9 | 4.5 | 15.3 | 11.9 | 12.9 | 8.6 | 6.8 | 7.1 | 11.3 | 11.7 | 11 | 5.9 | 5.9 | 5.7 |
| 10 | 10.4 | 10.8 | 12.2 | 5.4 | 5.9 | 6.8 | 11.3 | 11.3 | 10.4 | 5.9 | 5.4 | 5.4 | 11.3 | 13 | 14 | 6.3 | 7 | 8.3 | 10.8 | 11.4 | 12.4 | 5.4 | 5.8 | 6.1 |
| *Mean* | *12* | *11.26* | *11.27* | *6.14* | *5.99* | *6.28* | *10.92* | *10.02* | *9.66* | *5.76* | *5.48* | *5.3* | *12.46* | *13.21* | *12.92* | *7.62* | *7.63* | *7.65* | *11.32* | *11.67* | *11.73* | *5.95* | *5.91* | *6.08* |
| *SD* | *2.51* | *0.83* | *1.73* | *0.99* | *0.51* | *1.11* | *1.55* | *0.88* | *0.87* | *1.16* | *0.40* | *0.50* | *1.78* | *0.92* | *1.06* | *1.08* | *0.60* | *0.79* | *0.82* | *0.82* | *1.04* | *0.41* | *0.29* | *0.40* |

Table 7.1. A typical measurement example from Subject 1; R1-R3, repetitions; P1-10, peaks; SD, standard deviation.

Mean of Coefficient of variation (%)

| Site | Maximum Velocity | | Mean Velocity | |
|---|---|---|---|---|
| | Peaks | Repetitions | Peaks | Repetitions |
| lhtl | 12 | 5 | 12 | 7 |
| lhwl | 12 | 6 | 11 | 6 |
| rhtl | 10 | 6 | 10 | 6 |
| rhwl | 11 | 7 | 11 | 7 |

Table 7.2. Mean coefficient of variation of peaks and repetitions of maximum and mean velocity; lhtl, left hand thenar level; lhwl, left hand wrist level; rhtl, right hand thenar level.

Correlation and Paired t-test

| Site | Maximum Velocity | | | | Mean Velocity | | | |
|---|---|---|---|---|---|---|---|---|
| | Mean | SE | p | r | Mean | SE | p | r |
| lhtl | 0.2 | 0.33 | 0.54 | 0.81 | 0.41 | 0.16 | 0.02 | 0.82 |
| lhwl | -0.76 | 0.89 | 0.4 | 0.74 | 0.38 | 0.2 | 0.07 | 0.71 |
| rhtl | 1.31 | 1.14 | 0.26 | 0.67 | 0.18 | 0.25 | 0.49 | 0.6 |
| rhwl | 0.34 | 0.35 | 0.35 | 0.71 | 0.15 | 0.18 | 0.42 | 0.71 |

Table 7.3. Paired t-test of mean velocity differences; SE, standard error; p, statistical significance; r, correlation; lhtl, left hand thenar level; lhwl, left hand wrist level; rhtl, right hand thenar level; right hand wrist level.

was not significant. A higher FPL maximum velocity measurement may be expected at the dominant side because of a higher skill than that at the non-dominant side. In addition to thumb flexion/extension, the subjects were asked to reach the full range of motion without using the accompanying muscles. In an earlier unpublished study we observed that a higher maximum velocity of FPL was easier obtained when the thumb was flexed and extended with intermediate pauses rather than with continuous flexion/extension.

The peak force and power output (and thus, indirectly, tendon velocity) of a muscle depend on numerous factors[16] such as: a) muscle and fibre size and length, b) architecture, such as the angle and physical properties of the fibre-tendon attachment and the fibre to muscle length ratio, c) fibre type, d) number of cross-bridges in parallel, e) force per cross bridge, f) peak dF/dt (peak level of force development) N/s, g) force velocity relationship, h) fibre maximum velocity, i) force-p and calcium relationship, and j) the force-frequency relationship. It is clear that if some of these components change due to any reason, such as a trauma which affects the first and second factor, the power output and the shortening velocity will also be affected.

The fibre maximum velocity can be measured by a slack test in vitro, but the single muscle shortening velocity or single tendon excursion velocity has not been previously measured directly from the tendons in vivo. In pathological situations in which the factors mentioned above are changed, tendon velocity measurements may be valuable to evaluate the impaired muscle condition. During our experiments, the average FPL maximal displacement velocity was measured. These measurements are valuable for relative changes, such as the difference between healthy and pathologic extremity. The values obtained from the apparatus are not calibrated to real displacement velocity of FPL tendon. Since the exact software of the apparatus was not available, filters and post measurement calculations are unknown. It will be useful to compare in vivo measurements of maximum velocity the apparatus with phantom models in which tendon displacement velocity is precisely known and can be manipulated.

## Conclusion

This study is the first to demonstrate that the velocity of single muscle contraction can be measured non-invasively with CDI; the difference in FPL tendon velocity between the wrist and thenar levels was non-significant; ten repetitions of thumb flexion/extension deliver an accurate measurement of maximal FPL tendon velocity; FPL tendon velocity varies considerably between healthy subjects.

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# Chapter 8

This chapter is mainly based on the following publication

H.M. Buyruk, H.J. Stam, C.J. Snijders, H.A. Schut, J.S. Laméris, W.P.J Holland, T.H. Stijnen. Non-invasive Tendon Excursion Measurement with Colour Doppler Imaging: in Vivo Application of a New Technique (In preparation).

# Non-invasive Tendon Excursion Measurement with Colour Doppler Imaging: in vivo application of a new technique

## Chapter 8

**Abstract**
The assessment of tendon excursion in the hand and wrist region is common in hand surgery and rehabilitation. Current available non-invasive techniques are not accurate. Although invasive methods are more precise, they can not be repeated many times, and can only be used for experimental purposes. The use of Colour Doppler Imaging (CDI) in the musculoskeletal system has been addressed recently in the literature by our group. In our earlier studies CDI has been successfully used for imaging of tendon function and for measurement of tendon displacement velocity. The aim of the present study is to assess the applicability of tendon displacement measurement by means of CDI on healthy volunteers and patients with hand pathologies, and to present the preliminary results. Four healthy volunteers and eight patients were included in the study. Patients with different hand disorders were selected from our hand polyclinic in the department of rehabilitation. A Philips AD1 (Angio Dynagraph 1) with software version 2.5 pulsed multi-channel CDI scanner was used. Patients were measured only once bilaterally and the same protocol of measurements were repeated three times for healthy subjects. Variance analysis for flexion measurements showed that there is a highly significant difference between the subjects ($p<0.01$), no significant difference between the sides ($p=0.70$) and a significant difference between the tendons ($p=0.02$). The same test for extension measurements showed that there is a highly significant difference between the subjects ($p<0.01$), no significant difference between the sides ($p=0.85$) and no significant difference between the tendons ($p=0.10$). The Student's t-test displayed a significant difference between the healthy side and the restricted side ($p<0.01$). The main conclusions from this study are; tendon excursions expressed in millimetres can be measured non-invasively by CDI; large inter and intra individual finger flexion/extension patterns exist in healthy persons; in this preliminary study significant left-right differences were found in patients which met the clinical examination results.

**Keywords**
Doppler, tendon, excursion, hand, ultrasound

### Introduction

The assessment of tendon excursion in the hand and wrist region is common in hand surgery and rehabilitation. Formation of adhesions around the tendons after hand surgery and inflammations occur frequently. A minimal excursion value is important to prevent the formation of adhesions. Non-invasive imaging and quantification of tendon displacement is also important for the evaluation of the muscle function, for discrimination of adhesions from ruptures and muscle inactivity, for determination of relative left and right extremity difference, and for post hand-surgical follow-up. The present non-invasive method is based on the calculation of tendon displacement from the joint rotation angle and its radius[1,2,3]. The measurement of the fingertip palmar distance can also indirectly provide information about tendon displacement. These methods produce values that are not highly accurate because joint radius and joint angle measurements are not precise. In addition to this low accuracy, it is not possible to discriminate more than one tendon effecting one joint, like the proximal interphalangeal (PIP) joint. Invasive measurements are possible during surgery or with insertion of radiopacque markers in the tendon by means of X-ray imaging. Although invasive methods are more precise, they can not be repeated many times and can only be used for experimental purposes[4].

The use of Colour Doppler Imaging (CDI) in the musculoskeletal system has been addressed recently in the literature by our group[5]. In our earlier studies CDI has been successfully used for imaging of tendon function and for measurement of tendon displacement velocity[6,7,8]. CDI has the advantage of non-invasive quantification of a single tendon velocity. As a result of pilot measurements with CDI it was possible to obtain values for tendon excursion from the integration of continuous velocity

measurements. Validation of this technique has been performed on a plastic phantom model and on human specimens. The aim of the present study is to assess the applicability of tendon displacement measurement by means of CDI on healthy volunteers and patients with hand pathologies, and to present the preliminary results. To obtain absolute values calibrations on the phantom model are used[9].

**Materials and Methods**

Four healthy volunteers and eight patients were included in the study. The volunteers (aged between 22 and 33) had full hand function without history of primary and secondary hand pathology. Patients with different hand disorders were selected from our hand polyclinic in the department of rehabilitation. The main inclusion criterium was unilateral finger flexion range of motion (ROM) restriction. Exclusion criteria were absence of any finger flexion, open wounds, general condition of the patient, and non-cooperation (Table 8.1). Informed consent was given by the subjects and patients.

**Table 8.1.** Patients involved in the study

| | Age (yr) | Pathology | Tendon | Side |
|---|---|---|---|---|
| **Patient 1** | 20 | Short tendon | FDP4 | Right |
| **Patient 2** | 57 | Joint stiffness | FDP4 | Right |
| **Patient 3** | 42 | Adhesion | EPL | Right |
| **Patient 4** | 27 | Joint Stiffness | EPL | Left |
| **Patient 5** | 40 | Short muscle due to contraction | FDP4 | Left |
| **Patient 6** | 36 | Joint stiffness | FDP4 | Left |
| **Patient 7** | 29 | Joint Stiffness | EDC3 | Right |
| **Patient 8** | 12 | Adhesion | FPL | Right |

Since the aim of this preliminary report is to introduce the in vivo application of the CDI technique, measurements of the flexor pollicis longus (FPL), third flexor digitorum profundus (FDP) and flexor digitorum superficialis (FDS) tendons were performed on healthy subjects. The measurements on patients were done on the tendons with pathology and compared to the healthy side. Subjects were sitting with their arms resting on the examination table. CDI was placed on the volar side of the wrist proximal to the radiocarpal joint. After a series of pilot studies it appeared that maximal excursion of the FDS and FDP of a finger can be obtained when the other fingers are held in flexed position. The subjects were asked to continuously flex and extend as fast as possible with a pause of approximately one second between extension and flexion. Twenty flexion and extension movements were minimally performed while the ten fastest of them[7] were used for data analysis. Patients were measured only once bilaterally but the same protocol of measurements were repeated three times for healthy subjects.

*Experimental set-up (Figure 8.1):* A Philips ADI (Angio Dynagraph 1) with software version 2.5 pulsed multi-channel CDI scanner was used. A procedure similar to blood flow examinations was followed. A 7.5 MHz linear transducer and an 18° wedge between the transducer and the skin was applied. After a clear Doppler image of the excursion was obtained in the threshold mode, the apparatus was switched to the spectrum mode. The angle between the ultrasound transducer and the tendon was corrected with the angle correction option of the apparatus. The Doppler signals were transferred from the CDI apparatus to a bidirectional Doppler instrument (Versatone D9, Medsonics)[10]. The signal from this Doppler instrument was registered in a personal computer (PC) with an a/d converter (PCL 818L) and a data acquisition program

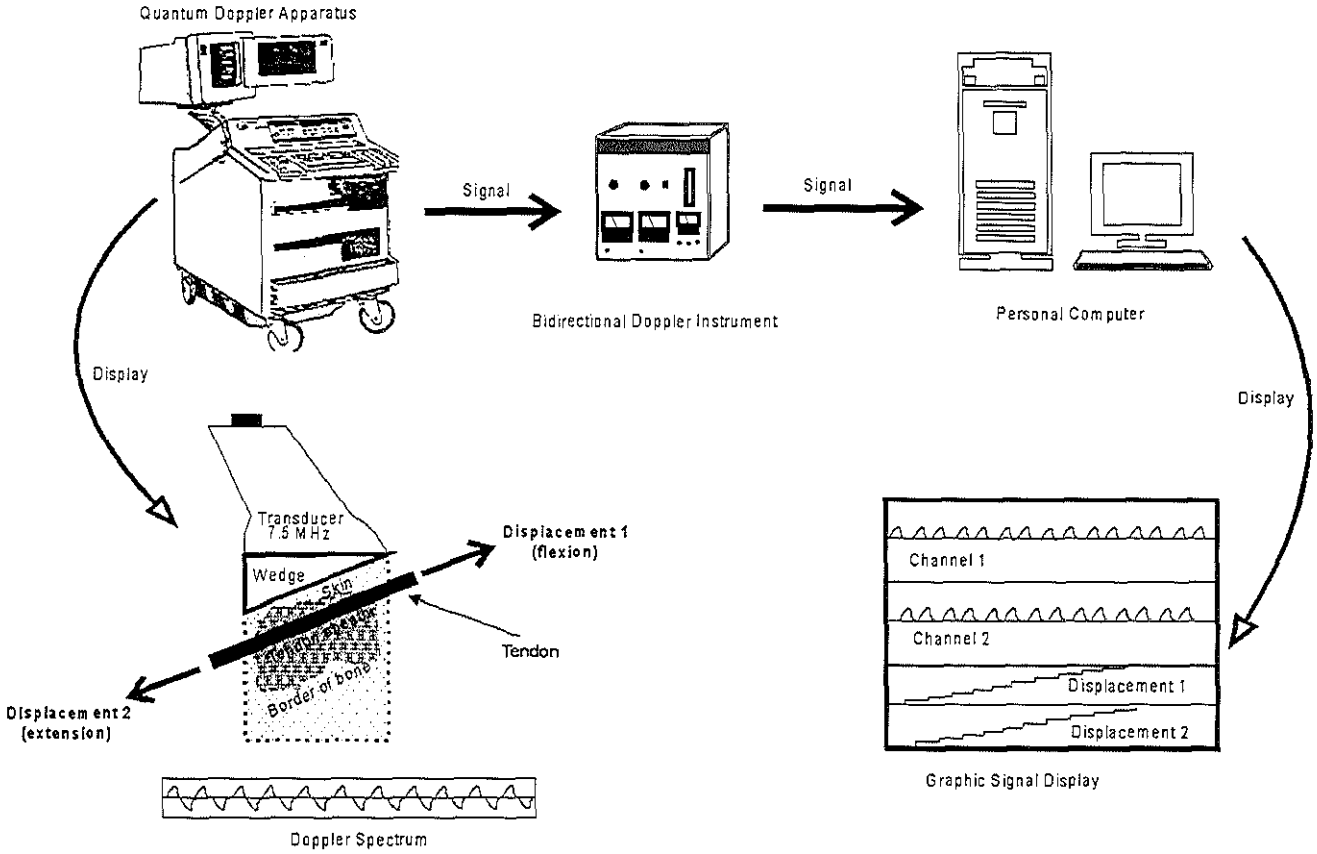


**Figure 8.1.** Experimental set-up.

(MKR, tailor-made). During the measurements Doppler waves are on line for flexion velocity, flexion excursion, extension velocity, extension excursion, and excursion differences between flexion and extension, in five graphic windows on the screen.

Analysis of variance was done to the data collected from the healthy subjects. The variability between the subjects, sides and tendons was analysed separately for extension and flexion. The data from left and right hand of the patients were compared with Student's t-test. For statistical significance a level of $p<0.05$ was chosen.

## Results

It was possible to measure tendon displacement with all the subjects and patients. Figure 8.2 is a typical example of formats of the PC monitor screen on line during measurements. In table 8.2 the measurement results from healthy subjects are given. Variance analysis for flexion measurements showed that there is a highly significant difference between the subjects ($p<0.01$), no significant difference between the sides ($p=0.70$) and a significant difference between the tendons ($p=0.02$). The same test for extension measurements showed that there is a highly significant difference between the subjects ($p<0.01$), no significant difference between the sides ($p=0.85$), and no significant difference between the tendons ($p=0.10$).

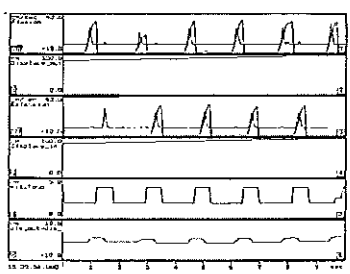

**Figure 8.2.** A typical example of formats of the PC monitor screen on line during measurements. Five windows on the PC monitor displayed on line flexion velocity, flexion excursion, extension velocity, extension excursion, and excursion difference between flexion and extension.

**Table 8.2. The measurement results from the healthy subjects**

| | | Tendons | | | | | |
|---|---|---|---|---|---|---|---|
| | | FDP3 | | FDS3 | | FPL | |
| | Sides | Flexion | Extension | Flexion | Extension | Flexion | Extension |
| Subject 1 | L | 1.40 | 1.23 | 1.55 | 1.42 | 1.41 | 1.39 |
| | R | 1.64 | 1.55 | 1.48 | 1.52 | 1.21 | 1.00 |
| Subject 2 | L | 1.56 | 1.52 | 1.37 | 1.20 | 1.68 | 1.62 |
| | R | 1.49 | 1.40 | 1.50 | 1.40 | 1.43 | 1.27 |
| Subject 3 | L | 0.56 | 0.57 | 0.58 | 0.50 | 0.34 | 0.32 |
| | R | 0.49 | 0.50 | 0.48 | 0.43 | 0.43 | 0.46 |
| Subject 4 | L | 0.81 | 0.76 | 0.43 | 0.41 | 0.50 | 0.50 |
| | R | 0.72 | 0.67 | 0.56 | 0.58 | 0.55 | 0.55 |

In figure 8.3 the mean of measurement results from patients are presented. The Student's t-test displayed a significant difference between the healthy side and the restricted side ($p<0.01$).

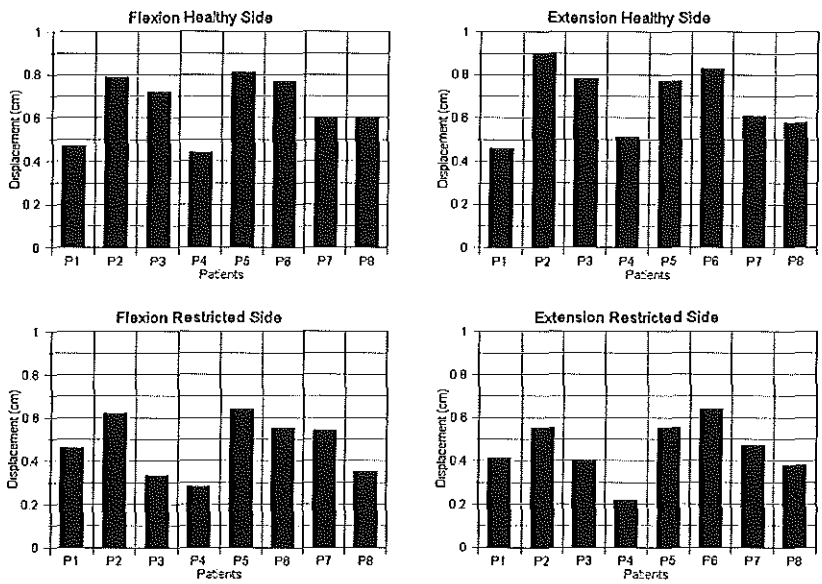


**Figure 8.3.** The bilateral measurement results of the patients

**Discussion**

An objective instrumented non-invasive measurement of tendon excursion is not previously reported in the literature. The availability of a non-invasive technique would be attractive for routine application in hand surgery and rehabilitation, as well as in orthopaedics and neurology. Tendon excursion can be affected by limited active range of motion and adhesions, but restricted displacements can also be related to muscle disfunction.

With the present study we could demonstrate that CDI can produce the in vivo displacement of tendons in the hand region expressed in millimetres. Large inter and intra individual flexion/extension pattern differences were observed in healthy persons. This indicates that future standardization is a prerequisite before clinical application of the method comes in routine. When moving the finger joints fast, patients tend to limit the active range of motion. This may be prevented by an audible feedback when reaching the limits of active extension and flexion. The audible feedback ensures movement through the full range of active motion. The intimate relation of flexors, especially FDS muscle bellies, and extensors, may limit active range of motion (AROM). Therefore positioning of the fingers not involved in the test is critical. Full extension of all the fingers limits flexion of the FDP tendon which is measured. In this respect we think of positioning the lower arm with a splint and selecting a repeatable motion pattern. Learning seems also to be an important factor. Healthy persons could be asked to repeat the test three times, however this appeared to be too strenuous to some patients. Fast flexion/extension movements are preferable since the CDI produces measurement results with less variation in calculated

displacement. For the same reason a pause of at least 1 second was introduced between every flexion/extension movement. This facilitates the discrimination of two subsequent Doppler peaks. The test person did not experience inconvenience from the measurements if they are limited to approximately 30 flexions and extensions. In accordance with the selection of patients with unilateral finger flexion restriction, we recorded significant left-right differences, which confirmed the clinical status of the respective individuals.

**Conclusions**

- Tendon excursions expressed in millimetres can be measured non-invasively by CDI;
- Large inter and intra individual finger flexion/extension patterns exist in healthy persons;
- In this preliminary study significant left-right differences were found in patients, this met the clinical examination.

**Acknowledgements**

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# Chapter 9

This chapter is mainly based on the following publication
W.P.J. Holland, H.M. Buyruk, E. Hoorn, C.J. Snijders, J.S. Laméris, H.J. Stam. A Simple Reciprocatingly Moving String Test Target for Calibration of Colour Doppler Displacement Measurements (in preparation).

# A Simple Reciprocatingly Moving String Test Target for Calibration of Colour Doppler Displacement Measurements

## Chapter 9

**Abstract**
Tendon excursions are frequently measured in clinics especially in hand surgery and rehabilitation. This article describes a simple phantom model to test Colour Doppler Systems on their ability to measure the movements of the tendon. The aim of this study is to compare Doppler measurements with values from a displacement metre which is directly connected to the string. In case differences are detected the aim is to develop and test the necessary filters and correction constants to improve the precision. The model consists of a rubber string moving in a tank filled with liquid. Flexion and extension of the tendon are simulated by moving the string manually. The string has been attached to a digital displacement metre for calibration of registered movements. A multi-channel pulsed Colour Doppler Imaging (CDI) is used in the experiments. Doppler signals are transferred first to an envelope detector and then to a PC where Doppler velocity signals are integrated to values of excursion of string. Mean and standard deviations of the displacement values in these data sets are estimated as 2.3 ± 0.3 cm/sec and 2.1 ± 0.3 cm/sec for flexion and extension respectively. These average values are between 20% and 15% respectively under the executed movement of 2.7 cm. Linear regression analysis yields slopes of 0.012 cm/(cm/sec) and 0.013 cm/(cm/sec) respectively, indicating only a slight dependence of measured displacement from peak velocity.

**Keywords**
ultrasound Doppler, string target, test phantom, tendon speed measurements

### Introduction

Tendon excursion measurements are frequently used in clinics especially in hand surgery and rehabilitation. Non-invasive measurement methods are based on goniometric calculations from finger joint angle and radius or on finger tip palmar distance measurements[1,2]. These measurements are rather unreliable and subjective. Invasive measurement methods are X-ray imaging of the tendons with insertion of radio-opaque materials or measurements during hand surgery operations[3]. These invasive methods are more reliable but are mostly used for experimental purposes.

Recently described methods by our group for imaging of muscle and tendon activity[4,5,6] and non-invasive measurement of the tendon excursion speed[7] is based on ultrasound Doppler speed measurement with a Colour pulsed Doppler Imaging (CDI) system. CDI combines a B-mode grey scale display with colour-coded movement information; it has been primarily developed to examine the blood flow stream in the arterial system[8]. A CDI system radiates ultrasound on both stationary and moving objects. The scattered sound received back from moving red blood cells and tissue structures are increased or decreased in frequency, depending on the magnitude and direction of the velocity of the moving objects. This so-called Doppler shift effect is then used to display the moving objects in red or blue on the screen of the CDI. With this method we use the demodulated Doppler spectral signals for determining the speed and excursion of a moving tendon. During the development and validation of this method it appeared desirable to have available a test phantom for simulating the movements of a tendon in the tissue.

Phantoms for the testing of CDI systems have been described in the literature, working with a moving string[9,10,11,12] or band[3]. All these phantoms are equipped with motors to generate constant or programmable pulsatile string speeds. A hand powered test phantom was considered suitable for simulating the different speeds of extension and flexion of a tendon in the body. The aim of this study is to compare Doppler displacement measurements with displacement metre values which are directly connected to the

string. If there are differences the aim is to develop and test the necessary filters and correction constants to improve the precision.

**Material and methods**

*Materials-* The test model (Figure 9.1) consists of a glued perspex tank (1) with the sizes of 20x8x8 cm. A round rubber string (2) (4mm diameter) with the ends connected and suspended by a spiral spring (3) is placed in a tract through six pulleys (4). The string runs outside the tank under the bottom and alongside the two sides, and through the tank at 2 cm above the bottom. The tank contains degassed tap water by vacuum till a level of 4 cm above the string. The part of the string under water runs through an echo poor plastic straw (5; 5 mm diameter, 12 cm length). The straw is kept in its place by four remote supports (6) connected to the wall. The straw restricts undesired movements of the string in transversal plane and simulated a tendon sheath. At the side of the model a small plastic block (7) is connected to the string which serves as a handle to let the string move. This block is connected during the movement via a metal plate (8) to the telescopic end of a digital displacement metre (9), type Mitutoyo IDU 25E with D/A interface and a maximal range of 2.70 cm. Therefore the to and fro movements of the string is restricted to the mechanical maximal range of the displacement metre.

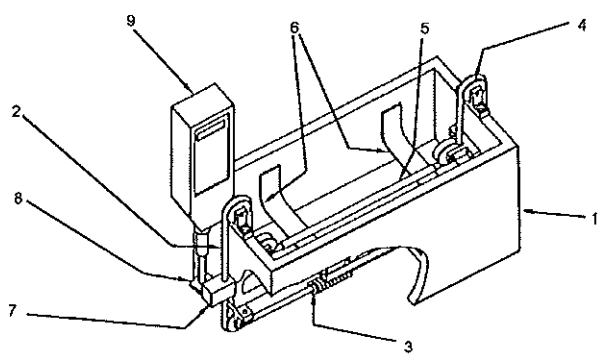


**Figure 9.1.** Drawing of the test phantom. Assigned components are mentioned in the text.

A multi-channel pulsed CDI, type Philips Quantum AD1 with software version 2.5 and upgrade kit option, is used in the experiments. The measurements were done with 7.5 MHz phased linear array transducer provided with an $18^0$ wedge angle. The transducer is fixed with a stand, positioned parallel to the string in the water to image the string running horizontally. The CDI worked in single point spectrum mode and the image of the string is selected with the assistance of a marker on the screen. By means of the trackball the marker is adjusted to the optimal position in order to obtain a maximal loud Doppler signal when the string is moved to and fro. The Doppler signals are collected via the stereo headphone audio plug exit on the apparatus. The scan depth is put at 6.7 cm (pulse repetition frequency PRF=10.48 kHz), and the medium flow level (aliasing frequency +/-2.6 kHz). Further settings were: transducer power -20dB, initial gain -65dB, slope of the gain 0dB/cm and a spectrum Doppler gain level of 20. With these settings non-distorted Doppler sound signals were obtained.

*Measurement procedure-* Experimental set-up is shown schematically in figure 9.2. The up and down movements of the string were executed manually from both sides of the tank manually. This two-folded

power increases the control on the movement of the string and decreases the generated unintended stretch of the string. Doppler signals were processed while the string was moved manually over the maximal displacement track of 2.7 cm. This was repeated, approximately 200 times with different excursion periods, the peak velocities varying between 4 and 30 cm/sec. In this way Doppler signals were generated, with peak frequencies between 120 - 1000 Hz.

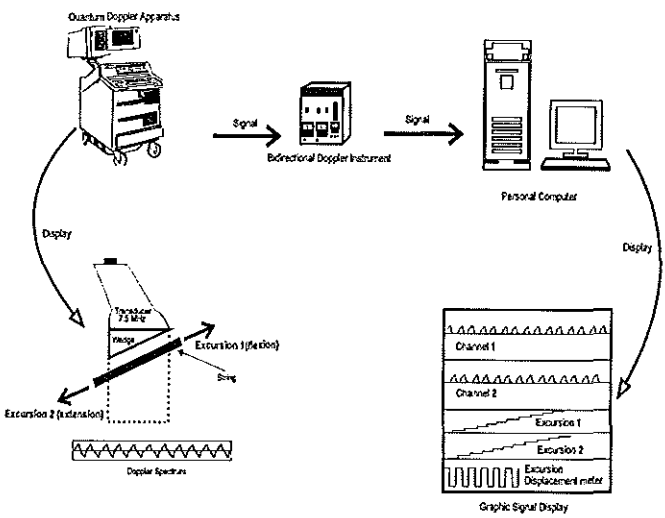


**Figure 9.2.** Experimental set-up.

*Signal processing*- The CDI system supplies the Doppler spectral signals which contain the speed-information for the to and fro movements. The two channel bidirectional Doppler signals are transferred to two external envelope detectors for frequency to voltage conversion for determining the average speed during flexion and extension. The electronic zero-crossing processors section of a bidirectional Doppler instrument (versatone model D9, Medsonics) is used for this purpose. The velocity-proportional signals at the output of the zero crossing processors and the output signal from the displacement metre are sampled and digitalized (25 samples/s) by a multi-channel A/D convertor and were subsequently processed by a data acquisition program (MKR, internal development) running on a PC. This program also converts on line the imported velocity-proportional data to speed data which are combined with the displacement metre data and saved as files on a hard disk. The conversion to speed ($v$) is made by the Doppler formula[8].

$$v = f_d * c/(2 * f_t * \cos(\theta))$$

In this formula $f_d$ is the Doppler frequency, $c$ is the propagation velocity of ultrasound (1500 and 1550 m/s for water and tissue, respectively), $f_t$ is the ultrasound-transmission frequency (7.5 Mhz for the transducer used), and $\theta$ is the angle between the moving string and the ultrasound bundle. The on line program uses a fixed angle of 72 degrees as a first estimate for calculating raw on line speed and on line excursion values. Calibration of the convertors and the algorithms of the on line program is accomplished by using a 1kHz sinus shaped signal as the Doppler signal. As expected from the Doppler formula, the calculated speed should then be 32.4 cm/sec or 33.6 cm/sec in the case of water and tissue respectively, as the propagation

medium. These values are used as calibration values in the on line software program.

The files obtained from a series of measurements are further processed through the special purpose off line program (SPD2DISP, tailor made). This program requires from the user to input the real Doppler angle as prevailing during the time of measurement. The program then performs exact Doppler angle corrections to obtain the final off line speed values, smooths the Doppler velocity waveforms through an optional moving average filter, segments velocity waveforms, and integrates the valid waveforms to off line displacement values for flexion and extension excursions. These values are then added to the content of the data files. The program also displays the average of the ten longest excursions, which serves as a feed back aid for quickly evaluating the in vivo measurements.

**Results**

It was possible to obtain displacement values for to and fro movements simulating flexion and extension. The obtained raw data was processed by the SPD2DISP program with real Doppler angle and a three point moving-average filter. Figure 9.3 presents a sample screen of registered velocity and displacement waves. The windows on this screen display in order from top to bottom: flexion velocity and off line excursion, on line flexion excursion, extension velocity and off line excursion, on line extension excursion, displacement metre excursion, and the difference between extension and flexion excursions. All valid data is saved as ASCII files separately for flexion (n=183) and extension (n=191) and is collected in the scatter diagrams of figure 9.4 and 9.5. In these diagrams the calculated Doppler displacements are placed against the corresponding velocity peaks. Mean and standard deviations of the displacement values in these data sets are estimated as 2.3 ± 0.3 cm/sec and 2.1 ± 0.3 cm/sec for flexion and extension respectively. These average values are between 20% and 15% respectively under the executed movement of 2.7 cm. Linear regression analysis yields slopes of 0.012 cm/(cm/sec) and 0.013 cm/(cm/sec) respectively, indicating only a slight dependence of measured displacement from peak velocity. These results can serve to correct future measurement results of tendon excursions.

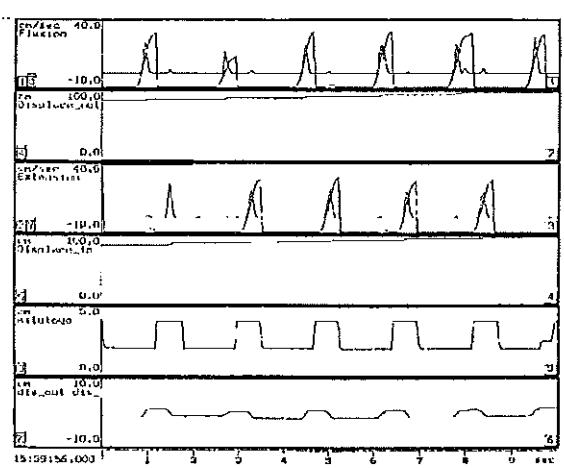

**Figure 9.3.** In five windows, speed and movements signals are shown as measured and calculated during the exercise of some to and fro movements of the string of the test phantom. Window 1: flexion speed and off line excursion; window 2: off line flexion excursion; window 3: extension speed and off line excursion; window 4: on line extension excursion; window 5: displacement metre excursion; window 6: the difference between flexion and extension excursions.

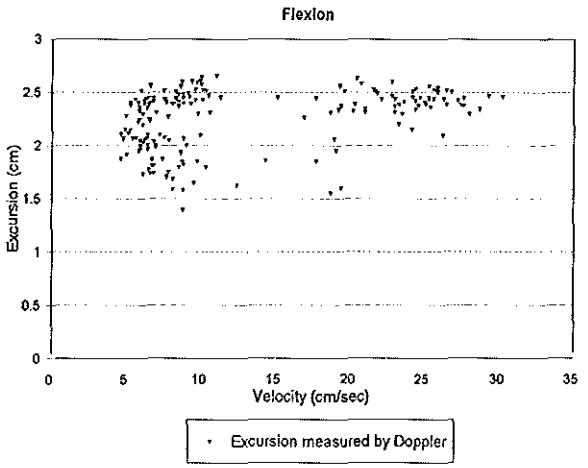


**Figure 9.4.** Scatter diagram of results of string extension excursion of 2.7 cm, executed at different speeds. Placed on the Doppler method determined speeds against the corresponding velocity peaks.

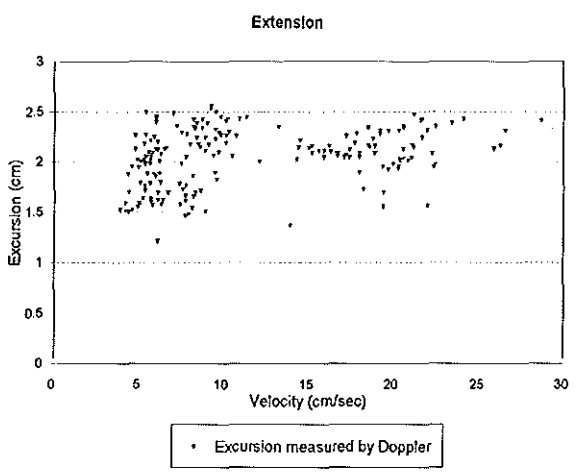


**Figure 9.5.** Scatter diagram of results of string flexion excursion of 2.7 cm, executed at different speeds. Placed on the Doppler method determined speeds against the corresponding velocity peaks.

## Discussion

The calibration procedures were satisfactory for excursion measurements with a CDI on a phantom model. Since the purpose of these measurements was to determine the excursions of the tendons especially in the hand and wrist region it was ideal to use a hand-powered phantom model. The variations and the range of the displacement speeds in this phantom model study will be the same in the future use of this method in the hand and wrist region. We therefore choose a simple phantom without motor, whereby a string is moved periodically by hand between two fixed end points. With this test phantom experiment the values showing

the amount of movement obtained with the Doppler can be directly compared with the values of executed string movements which are precisely known.

The deviation between the Doppler measurements and displacement metre measurements showed that the calculated calibration values have to be corrected. First possible cause of this deviation would be the excursion difference between the part of the string where the movement was initiated and the part of the string under the transducer. This would be caused by the extension of the string under load, therefore a control study was done on the model. The excursion of the string under the transducer is measured while the string was moved manually as usual. The result was 2.7±0.1 cm, which excludes the string extension as a cause for deviations. Second possible cause would be related to the type of envelope processor used here to average the frequency from the Doppler spectrum. Ideal would be a processor type which determines the true average frequency of the Doppler spectrum. However, we used a zero-crossing processor which output is proportional to the root means square (RMS) frequency of the input[8]. Since the spectrum obtained from the excursion of a string is rather narrow band it is not expected that the measurement results will differ much from both type of processors. The third possible cause could lie in the high-pass filters (wall filters), which are built in the CDI to suppress the strong low-frequency signals originated from non-moving or very slow moving tissues. If (low) frequencies from the desired movements are suppressed, measurement errors can occur. The cut-off frequency of the high-pass filter in the CDI (medium flow level) is 100 Hz. The Doppler frequencies of the longitudinal excursions are in the range of 120-1000 Hz., they are transmitted without attenuation, hence the conclusion is that the high-pass filters would not be the cause of measurement errors.

Consequently, we believe that an in vitro study on human specimens would deliver extra-information and supplemental control for our calibration values.

## Conclusions

- It was possible to measure string excursions with high correlation values but with a certain percentage of lower values;
- The speed of the excursion does not seem to influence the precision of the Doppler excursion measurements.


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# Chapter 10

This chapter is mainly based on the following publication

H.M. Buyruk, W.P.J. Holland, C.J. Snijders, J.S. Laméris, E. Hoorn, R. Stoeckart, H.J. Stam. Tendon Excursion Measurements with Colour Doppler Imaging: a calibration study on embalmed human specimens (Submitted to J Rehabilitation Sciences, 1996).

# Tendon Excursion Measurements with Colour Doppler Imaging: a calibration study on embalmed human specimens

## Chapter 10

**Abstract**
Assessment of tendon excursion is important, especially after tendon release operations in hand surgery and in rehabilitation. The use of Colour Doppler Imaging (CDI) in the musculoskeletal system has been addressed recently in the literature by our group. The aim of the present study is to assess the applicability of tendon displacement measurement by means of CDI on human specimens, which forms a media for measurements closer to in vivo conditions than a phantom model, and to assess the correlation between values measured by Doppler and by displacement metres. Experiments were performed on an embalmed human arm. Flexor pollicis longus (FPL), flexor digitorum superficialis three (FDS3) and flexor digitorum profundus two and three (FDP2 and FDP3) muscles were separately connected to a weight of 10 N with a steel wire running over a pulley. This weight moved the telescopic end of a digital displacement metre up and down during manually extended and flexed fingers of the specimen. Displacement measurements with Doppler were done with pulsed multi-channel CDI scanner. Doppler signals are transferred first to an envelope detector and then to a PC where Doppler velocity signals are integrated to values of tendon excursion. Statistically Doppler measurements were 3% lower, $p<0.00$ for both flexion and extension. CDI measurements of finger tendon excursion correlates highly with a mechanical micro displacement metre, the latter being considered a golden standard. Tendon excursion measurements with CDI on a human hand and arm give accurate results. Calibration values that are used in the software to convert CDI signals to tendon displacement values in vivo deliver a good correlation with the golden standard as used in the cadaver experiments.

**Keywords**
Doppler, tendon, excursion, hand, ultrasound, human specimens

### Introduction

Assessment of tendon excursion is important, especially after tendon release operations in hand surgery and in rehabilitation. The existing non-invasive methods such as excursion calculations from joint rotation angles and joint diameters are not precise[1,2,3]. Invasive measurement methods such as X-ray imaging of the tendons with insertion of radio-opaque materials or measurements during hand surgery are performed for experimental reasons[4].

The use of Colour Doppler Imaging (CDI) in the musculoskeletal system has been addressed recently in the literature by our group[5,6,7]. In our earlier studies CDI has been successfully used for imaging of tendon function and for measurement of tendon displacement velocity[8]. As demonstrated with pilot measurements with CDI it is possible to obtain values for tendon excursion from the integration of continuous velocity measurements. Validation of this technique has been performed on a plastic phantom model, and in vivo application has been presented in a preliminary report[9,10]. The aim of the present study is to assess the applicability of tendon displacement measurement by means of CDI on human specimens, which forms a media for measurements closer to in vivo conditions than a string phantom model, and to assess the correlation between values measured by Doppler and by displacement metres.

### Material and methods

Experiments were performed on an embalmed male human arm (age 87). The lower arm was dissected carefully 10 cm proximal to the radiocarpal joint from superficial to deep layers keeping

distal structures intact. All the muscles were identified and separated from each other, including the bellies of the superficial and deep digital flexor muscles two to five. Flexor pollicis longus (FPL), flexor digitorum superficialis three (FDS3) and flexor digitorum profundus two and three (FDP2 and FDP3) muscles were separately connected to a weight of 10 N with a steel wire running over a pulley. This weight moved the telescopic end of a digital displacement metre up and down (Mitutoyo IDU 25E with D/A interface and a maximal range of 2.70 cm) during manually extended and flexed fingers of the specimen. Displacement measurements with Doppler were done with a Philips Angio Dynagraph 1 (AD1) with software version 2.5 pulsed multi-channel CDI scanner. A procedure similar to earlier studies was followed[4,5,6,7](figure 10.1). A 7.5 MHz linear transducer and an° 18 wedge between the transducer and the skin was applied. After a clear Doppler image of the excursion was obtained in the threshold mode, the apparatus was switched to the spectrum mode. The angle between the ultrasound transducer and the tendon was corrected with the angle correction option of the apparatus. The Doppler signals were transferred from the CDI apparatus to a bidirectional Doppler instrument (Versatone D9, Medsonics). The signal from this Doppler instrument was registered in a personal computer (PC) with an a/d converter (PCL 818L) and a data acquisition program (MKR, tailor-made). According to the Doppler formula, the calculated speed should be 32.4 cm/sec or 33.6 cm/sec in the case of water and tissue respectively, as the propagation medium[11]. We used water propagation speed in the on line and off line programs[9]. During the measurements waves of flexion velocity, flexion excursion, extension velocity, extension excursion, and excursion differences between flexion and extension were seen on the screen on line in five graphic windows.

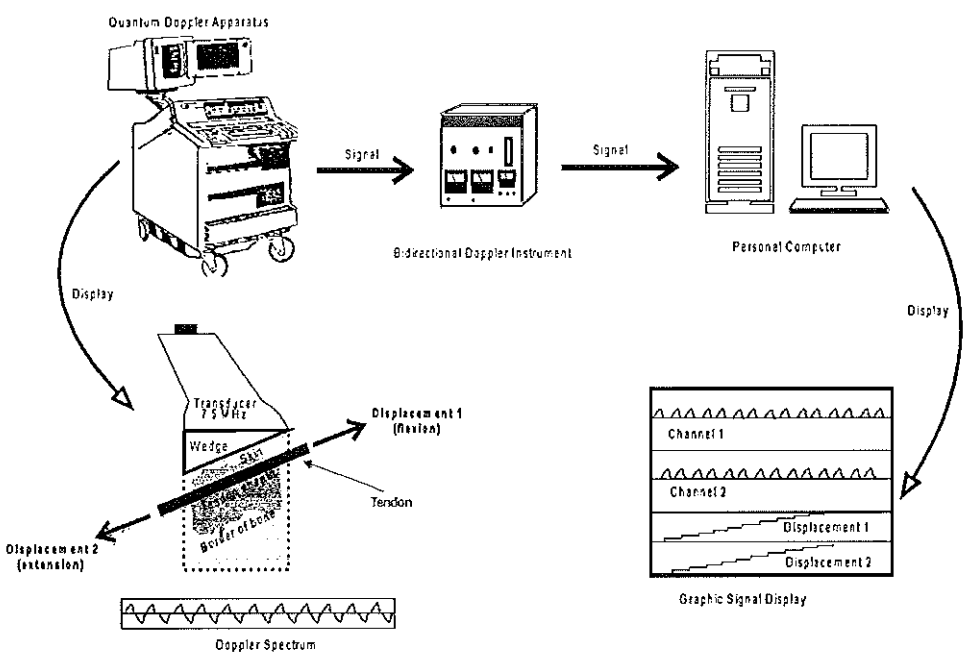


**Figure 10.1.** The experimental set-up

**Table 10.1. The cross table of Doppler extension measurements with displacement metre measurements**

| | | cm | | | | | | | |
|---|---|---|---|---|---|---|---|---|---|
| | | 0.5 | 0.6 | 0.7 | 0.8 | 0.9 | 1.0 | 1.1 | Total Count |
| | 0.4 | | 1 | | | | | | 1 |
| | 0.5 | 2 | | | | | | | 2 |
| | 0.6 | 1 | 15 | 7 | 1 | 1 | | | 25 |
| cm | 0.7 | | 1 | 22 | 16 | 5 | 1 | | 45 |
| | 0.8 | | 1 | 3 | 30 | 8 | 1 | | 43 |
| | 0.9 | | | 1 | 3 | 14 | 3 | 1 | 22 |
| | 1.0 | | | | | 5 | | | 5 |
| | 1.1 | | | | | | 1 | | 1 |
| Total Count | | 3 | 18 | 33 | 50 | 33 | 6 | 1 | 144 |

**Table 10.2. The cross table of Doppler flexion measurements with displacement metre measurements**

| | | cm | | | | | | |
|---|---|---|---|---|---|---|---|---|
| | | 0.5 | 0.6 | 0.7 | 0.8 | 0.9 | 1.0 | Total Count |
| | 0.5 | 2 | | 1 | | | | 3 |
| | 0.6 | | 10 | 10 | 3 | | | 23 |
| | 0.7 | 1 | 2 | 21 | 15 | 1 | | 40 |
| cm | 0.8 | | 2 | 2 | 27 | 7 | | 38 |
| | 0.9 | | | 1 | 6 | 13 | 2 | 22 |
| | 1.0 | | | | 1 | 9 | 5 | 15 |
| | 1.1 | | | | | | 3 | 3 |
| Total Count | | 3 | 14 | 35 | 52 | 30 | 10 | 144 |

Each muscle (FPL, FDS3, FDP2 and FDP3) displaces together with the manually powered flexion and extension of the corresponding finger. The flexion and extension of the fingers are executed with irregular ranges of motion including MCP, PIP and DIP joints. Approximately 20 extensions and flexions were recorded from each muscle in every session and this procedure was repeated three times.

The transducer was placed between the radiocarpal joint and the dissection line, and above the anatomical location of the measured tendon. From the on line program files, the first 12 extension and flexion peaks were used for statistical evaluation. The t-test for paired samples was carried out for each tendon separately for flexion and extension. For statistical significance a level of $p<0.05$ was chosen.

**Results**

Since the muscles displayed an irregular range of displacements we used the ratios between the displacement metre and Doppler measurements for presentation of the results and statistics. As shown in table 10.1 and 10.2 the cross tables between the direct displacement measurements in cm and the Doppler integration values were good. Statistically Doppler measurements were 3% lower, $p<0.00$ for both flexion and extension. In figure 10.2 the excursion of the different tendons are presented separately, i.e. the average values resulting from manually exerted flexion and extension movements of the fingers. Since the FDP and FDS muscles were separated to four for each finger these values have no meaning for the range of motion, but are shown for visual comparison of average values obtained with the two measurement methods.

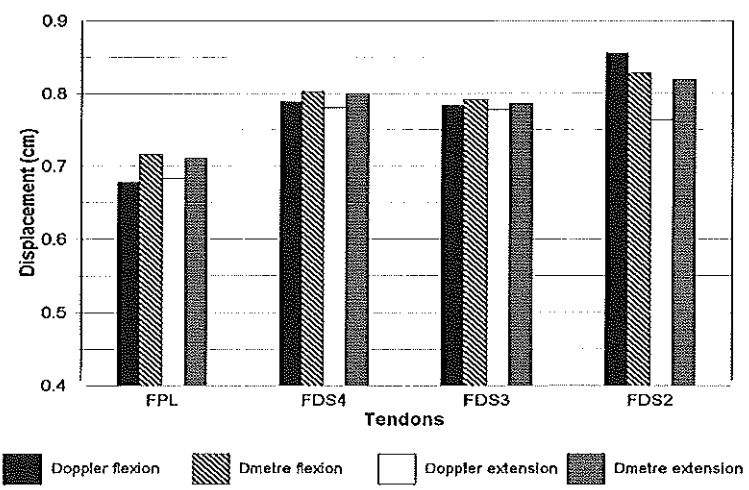


**Figure 10.2.** The average excursion of the tendons measured by Doppler and displacement metre during extension and flexion.

**Discussion**

It was possible to accurately convert CDI signals obtained from displacement of a string into millimetre excursion. In a previous study a phantom model made of plastic was used for this purpose, simulating tendon excursion in the human body. The calibration values which are mathematically calculated and tested on a string phantom model were further used in in vivo preliminary studies on tendon excursion in the hand and wrist area of healthy subjects and patients. Justification of this in vivo application of mentioned calibration values is difficult to prove, since a golden standard for in vivo tendon excursion is impossible non-invasively. Hence, the in vivo situation is approximated with experiments on human specimens and the golden standard was realized by means of mechanical micro displacement metres connected to the tendon. This set-up is closer to an in vivo situation than

the phantom model because the medium between probe and tendon is soft tissue instead of water and the reflection of ultrasound on tendon and string material is different. Variable ranges of motion of finger flexion and extension were executed intentionally because different excursions may affect the correlation between the different methods of measurement due to different filters which are effective at different speeds. The results of this validation study were not as expected, i.e. the correlation between the integrated Doppler signals and the golden standard was better than in the phantom experiments[10]. Calibration of the convertors and the algorithms of the on line program was accomplished by using a 1kHz sinus shaped signal as the Doppler signal. According to the Doppler formula, the calculated speed should then be 32.4 cm/sec or 33.6 cm/sec in the case of water and tissue respectively, as the propagation medium. There was also a 3% diffference between the water and tissue propagation speeds which corresponds with the results of this study. Since we used the water propagation speed values in this study it explains the difference appeared between the displacement metre and Doppler measurements. We ascribe this to the fact that the software in the CDI is tuned to soft tissue media. Because the converted CDI values were 3% lower than the golden standard, a corresponding correction will be applied on future in vivo measurements. We still do not know in the present stage what correction should be made on the calibration values for in vivo application. However, after the present study we assume that a correction value of 3% is more realistic than the 10% found in the phantom model study.

## Conclusions

- CDI measurements of finger tendon excursion correlates highly with a mechanical micro displacement metre, the latter being considered a golden standard;
- Tendon excursion measurements with CDI on a human hand and arm give accurate results;
- Calibration values used in the software to convert CDI signals to tendon displacement values in vivo deliver a good correlation with the golden standard as used in the cadaver experiments.

## Acknowledgements

We would like to express our thanks to I. Buyruk for her valuable contributions.

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# Summary

In the department of Rehabilitation research on and clinical examination of low back and hand pathologies are important. In both areas quantitative assessments for diagnosis, treatment and follow-up cannot be done with existing routine techniques.

For routine clinical examination and laboratory investigation many means are available. In this thesis the new applications for Colour Doppler Imaging (CDI) in the musculoskeletal system are investigated, specifically in the low back area and in the hand and wrist region. The study on low back pain raised special interest for the mechanical properties of the pelvic joints, in particular the sacroiliac (SI) joints. For the assessment of SI joint stiffness no instrumented method was available. The study on the hand and wrist pathology resulted in a new diagnostic method to measure non-invasively the function of the tendons, which had not been available up till then. While searching for solutions we discovered that CDI, which is normally used for blood flow examinations, opened new possibilities for the research of the musculoskeletal system in general, and our research interests in particular. CDI application on the assessment of SI joint stiffness, and hand and wrist pathologies is introduced in separate sectors which give a brief overview of the chapters of this thesis. The general aim of this thesis is to extend the application of CDI in the field of the musculoskeletal system. Ten studies, designed to validate the new applications of CDI, were done.

Study one investigated the behaviour of mass-spring systems on a pelvis model under vibrations, and assessed the effect of different artificially created stiffnesses on bilateral SI joints. The aim of this study was to demonstrate the proportional relation between joint stiffness and transmission of vibrations through the SI joints. A mechanical model of the pelvis was developed which represents the shape of the pelvic bones, the properties of the joints and of the major ligaments. While simulating a person lying in supine position, unilateral excitation of the pelvis at the anterior superior iliac spine generated vibrations which were received at the dorsal surface of the homo-lateral ilium and the sacrum by two accelerometers. Four different stiffness levels of the SI joints were combined as ten different situations, by means of metal plates, adjustable screws, and by tensions applied on artificial ligaments. The occurrence of resonance peaks in the ten simulated stability situations were evaluated in the frequency range between 25 and 600 Hz. The results showed that different SI joint stiffness levels in a mechanical pelvis model result in different resonance frequencies and transmission of vibrations across the SI joints. The transmission of vibrations through the SI joint was proportional to joint stiffness. The stiffness of the contra lateral joint influenced the dynamics of the joint on the measured side.

Study two dealt with the validity and reproducibility of an instrumented dynamic examination method to measure SI joint stiffness which was tested in vitro. In this study, four embalmed human female pelvises were excitated by a pelvic vibrator. A CDI scanner was used to image the amplitude of vibrations at different sites of the pelvis. Vibrations were applied to the anterior superior iliac spines unilaterally and were received by CDI all over the ipsilateral SI region. Three different stability conditions were created in the SI joints: no intervention, screwed, and with ligaments cut. Test results were quantified by taking the minimum threshold levels of the bones. The relative difference of vibration intensity between ipsilateral ilium and sacrum at each stability condition was accepted as the stiffness level for the SI joint. Statistics showed high reproducibility and significant differences between the stability conditions. Dynamic testing based on the use of vibrations provides visible and quantifiable intra- and inter-individual differences between SI joint stiffnesses. This new method is objective and reproducible. Future in vivo application is promising since there are no technical or safety restrictions.

Study three dealt with primary peripartum pelvic and low back pain which is a common complaint of females. The etiologic relation between pain and pelvic stability has been shown in previous studies, but at present there is no objective clinical testing method to evaluate pelvic stability. In this study, a dynamic measurement method using sonoelasticity to assess the SI joint stiffness was tested in vivo in 14 healthy female volunteers. With the subjects in supine position, vibrations were unilaterally

applied to the anterior iliac spine. The vibrations were registered by a CDI transducer over the ipsilateral SI joint. Since the threshold level of the apparatus is directly related to the power of the vibrations the intensity of the vibrations (sonoelasticity) on the sacrum and ilium was measured indirectly in threshold units. The differences between the threshold values were accepted as the power loss of vibrations through the SI joint. One-way Analysis of Variance-test and T-test for Paired Samples were applied on the measurement results ($P<0.05$). Statistically, the results showed a satisfactory intra-individual reproducibility and inter-individual variability. There was no significant difference between the data derived from the left SI joint and right SI joint. Based on the promising results on healthy female volunteers, this method will be specifically used in future studies on patients with peripartum pelvic pain (PPPP).

In study four the differences of SI joint stiffnesses between PPPP patients and healthy controls were compared with Doppler Imaging of Vibrations (DIV). PPPP is a condition related to the pathology of SI joints after the exclusion of other causes, such as lumbar disc hernia or spondylolisthesis. Although routine physical examination and radiologic investigations (X-rays, computerized tomography and magnetic resonance imaging) provide information about the pathology of the lumbar spine and the pelvis, these techniques do not contribute to the understanding of abnormal biomechanics. In previous studies we introduced a new technique to assess SI joint stiffness using CDI and vibrations, which is called DIV. This technique was first tested on a plastic model, embalmed human pelvises, and healthy volunteers. In this study the measurements were done on a group of PPPP patients ($n= 56$) and a healthy control group ($n= 45$). A vibrator, Derritron VP3 and a Colour Doppler Imaging apparatus (Philips Quantum AD1) were used. A protocol similar to that published earlier by our group was followed. The difference in SI joint stiffness between the patient group and the control group were statistically tested by means of the Wilcoxon's two sample test, the chi-square test and Student's t-tests. Both patients and healthy controls displayed hypo- and as well as hyper mobility, but there was no significant difference between these groups. However, there was a significant difference between the groups with regard to the average difference of SI joint stiffness between left and right. Asymmetric stiffness of the SI joints seemed to be more directly related to PPPP than the stiffness level of a single SI joint.

In study five the movements of the tendons and muscles of the hand and wrist region of the healthy volunteers were imaged with CDI. Most imaging techniques used to examine the musculoskeletal system, such as X-ray, CT scan, and MRI, produce mainly static images. CDI may offer the opportunity to study the dynamic characteristics of tendons and muscles. The purpose of this study was to explore the possibilities of the application of CDI in the imaging of the musculoskeletal system. A Philips Quantum Doppler was used to study the functional anatomy of the musculoskeletal systems of healthy subjects. This preliminary report on colour Doppler examination is focused on the imaging of wrist and hand. The results indicate that CDI offers a promising approach to image dynamically the tendons and muscles in the musculoskeletal system.

The purpose of study six was to examine the unilateral flexor and extensor tendon pathologies of the hand and compare these images with the healthy side. A Philips Quantum Doppler was used to study patients with tendon adhesions and trigger finger. The normal and pathological tendons were scanned at three anatomical levels; wrist, metacarpal, and proximal phalanx. The results obtained from the comparison of pathological images with normal ones indicated that CDI offers a promising approach to image dynamically and, in the future, to quantify the function of the tendons and muscles in the musculoskeletal system and to define their pathologies.

The purpose of study seven was to use the CDI for the measurement of maximum and mean tendon velocity. Recent studies showed that CDI can be used for the imaging of tendons in the hand and wrist region. Although other modalities are available for imaging of the musculoskeletal system, in vivo measurements of the velocity of tendon excursion are not possible. The flexor pollicis longus (FPL) tendon of 16 healthy volunteers was measured bilaterally at two levels (wrist and thenar). A splint from the fingers along the proximal lower arm was applied. The thumb was fixed to the splint from the first phalanx to allow flexion of the interphalangeal (IP) joint only. Pulsed CDI was used for the measurements. The maximum and mean velocities of the FPL tendon were measured at a spectrum

display mode during continuous voluntary contractions. At least ten sequential Doppler peaks (cm/sec) were recorded at every trial. The measurements were repeated three times. Paired t-test was applied and correlation coefficients were calculated between levels on the same side and the opposite side. No significant differences were found between the two levels of the same hand and those of the opposite hand. As expected, the data revealed variations in the inter-individual tendon velocities. The velocity of the excursion of the FPL tendon can be measured and reproduced well with CDI. It is expected that velocity measurements can be used in the future for the assessment of other tendons affected by various disorders.

The study eight was related to measurement of tendon excursions with CDI. The assessment of tendon excursion in the hand and wrist region is common in hand surgery and rehabilitation. Current available non-invasive techniques are not accurate. Although invasive methods are more precise, they can not be repeated many times, and can only be used for experimental purposes. The use of Colour Doppler Imaging (CDI) in the musculoskeletal system has been addressed recently in the literature by our group. In our earlier studies CDI has been successfully used for imaging of tendon function and for measurement of tendon displacement velocity. The aim of the present study is to assess the applicability of tendon displacement measurement by means of CDI on healthy volunteers and patients with hand pathologies, and to present the preliminary results. Four healthy volunteers and eight patients were included in the study. Patients with different hand disorders were selected from our hand polyclinic in the department of rehabilitation. A Philips AD1 (Angio Dynagraph 1) with software version 2.5 pulsed multi-channel CDI scanner was used. Patients were measured only once bilaterally and the same protocol of measurements were repeated three times for healthy subjects. Variance analysis for flexion measurements showed that there is a highly significant difference between the subjects ($p<0.01$), no significant difference between the sides ($p=0.70$) and a significant difference between the tendons ($p=0.02$). The same test for extension measurements showed that there is a highly significant difference between the subjects ($p<0.01$), no significant difference between the sides ($p=0.85$) and no significant difference between the tendons ($p=0.10$). The Student's t-test displayed a significant difference between the healthy side and the restricted side ($p<0.01$). The main conclusions from this study are; tendon excursions expressed in millimetres can be measured non-invasively by CDI; large inter and intra individual finger flexion/extension patterns exist in healthy persons; in this preliminary study significant left-right differences were found in patients which met the clinical examination results.

The study nine describes a simple phantom model to test Colour Doppler Systems on their ability to measure the movements of the tendon. The aim of this study is to compare Doppler measurements with values from a displacement metre which is directly connected to the string. In case differences are detected the aim is to develop and test the necessary filters and correction constants to improve the precision. The model consists of a rubber string moving in a tank filled with liquid. Flexion and extension of the tendon are simulated by moving the string manually. The string has been attached to a digital displacement meter for calibration of registered movements. A multi-channel pulsed Colour Doppler Imaging (CDI) is used in the experiments. Doppler signals are transferred first to an envelope detector and then to a PC where Doppler velocity signals are integrated to values of excursion of string. Mean and standard deviations of the displacement values in these data sets are estimated as $2.3 \pm 0.3$ cm/sec and $2.1 \pm 0.3$ cm/sec for flexion and extension respectively. These average values are between 20% and 15% respectively under the executed movement of 2.7 cm. Linear regression analysis yields slopes of 0.012 cm/(cm/sec) and 0.013 cm/(cm/sec) respectively, indicating only a slight dependence of measured displacement from peak velocity.

The purpose of the tenth study is to assess the applicability of tendon displacement measurement by means of CDI on human specimens, which forms a media for measurements closer to in vivo conditions than a phantom model, and to assess the correlation between values measured by Doppler and by displacement metres. Experiments were performed on an embalmed human arm. Flexor pollicis longus (FPL), flexor digitorum superficialis three (FDS3) and flexor digitorum profundus two and three (FDP2 and FDP3) muscles were separately connected to a weight of 10 N with a steel wire running over a pulley. This weight moved the telescopic end of a digital displacement meter up and down during

manually extended and flexed fingers of the specimen. Displacement measurements with Doppler were done with pulsed multi-channel CDI scanner. Doppler signals are transferred first to an envelope processor and then to a PC where Doppler velocity signals are integrated to values of tendon excursion. Statistically Doppler measurements were 3% lower, $p<0.00$ for both flexion and extension. CDI measurements of finger tendon excursion correlates highly with a mechanical micro displacement metre, the latter being considered a golden standard. Tendon excursion measurements with CDI on a human hand and arm give accurate results. Calibration values that are used in the software to convert CDI signals to tendon displacement values in vivo deliver a good correlation with the golden standard as used in the cadaver experiments.

The studies with CDI described above presented important possibilities for the diagnosis of musculoskeletal system disorders. Some technical properties of CDI technique seemed unexpectedly promising in this system, such as quantification of tendon and muscle excursions. During the validation studies of this technique on healthy subjects and patients it was also possible biomechanical aspects of low back and pelvic pain are found. This was only possible with the introduction of this new technique, Doppler Imaging of Vibrations (DIV).

# Samenvatting

In de Vakgroep Revalidatie is research en klinisch onderzoek van de lage rug en hand aandoeningen belangrijk. In beide gevallen kan de kwantitatieve bepaling voor diagnose, behandeling en de follow-up niet worden gedaan met de bestaande routine technieken.

Voor klinisch onderzoek en laboratorium onderzoek bestaan vele mogelijkheden. In dit proefschrift zijn de nieuwe toepassingen van Colour Doppler Imaging (CDI) in het bewegingsapparaat onderzocht, met name voor de lage rug, de hand en de pols. Tijdens de studie naar lage rugpijn ontwikkelde zich een speciale interesse naar de mechanische verhoudingen van de bekkengewrichten, met name de sacroiliac (SI) gewrichten. Voor de bepaling van de stijfheid van SI gewrichten was geen methode beschikbaar. De studie naar de hand en pols aandoeningen resulteerde in een methode voor het non-invasief meten van de functie van de pezen welke tot dan toe nog niet bestond. Tijdens het zoeken naar oplossingen ontdekten we dat CDI, welke normaal gebruikt wordt voor bloed-stroom onderzoek, nieuwe mogelijkheden bood voor onderzoek van het bewegingsapparaat in het algemeen, en voor onze onderzoeksinteressegebieden in het bijzonder. De toepassing van CDI voor het bepalen van de stijfheid van het SI gewricht en de hand en pols aandoeningen wordt geïntroduceerd in aparte alinea's welke een kort overzicht geven van de hoofdstukken van dit proefschrift. Het algemene doel van dit proefschrift is de uitbreiding van de toepassingen van CDI op het gebied van het bewegingsapparaat. Er zijn tien studies gedaan welke gericht waren op het valideren van de nieuwe toepassingen van CDI.

Studie één onderzocht het gedrag van de massa-veer systemen op een bekken model onder trillingen, en stelde het effect vast van de verschillende kunstmatig gecreëerde stijfheden van bilaterale SI gewrichten. Het doel van deze studie was om de proportionele relatie aan te tonen tussen stijfheid van het gewricht en overbrenging van trillingen door de SI gewrichten. Er werd een mechanisch model van het bekken ontwikkeld welke de vorm van de bekkenbeenderen weergaf, alsmede de verhoudingen van de gewrichten en dat van de belangrijkste ligamenten. Tijdens simulering van een persoon in rugligging veroorzaakte unilaterale excitatie van het bekken op de spina iliaca anterior superior trillingen welke werden ontvangen aan de dorsale oppervlakte van de homo-laterale ilium en het sacrum door twee versnellingsopnemers. Vier verschillende niveaus van stijfheid van de SI gewrichten werden gecombineerd tot tien verschillende situaties, door middel van metalen platen, verstelbare schroeven en door trillingen welke werden toegepast op de kunstmatige ligamenten. Het verschijnen van resonantie pieken in de tien gesimuleerde stabiliteit situaties werd geëvalueerd in een frequentie variërend van 25 tot 600 Hz. De resultaten toonden aan dat verschillende niveaus van stijfheid van het SI gewricht in een mechanisch bekken model resulteren in verschillende resonantie frequenties en transmissies van trillingen door de SI gewrichten. De transmissie van trillingen door het SI gewricht was proportioneel ten opzichte van de stijfheid van het gewricht. De stijfheid van het contra laterale gewricht beïnvloedde de dynamieken van het gewricht aan de gemeten zijde.

Studie twee onderzocht de validiteit en herhaalbaarheid van een geïnstrumenteerde dynamische onderzoeksmethode om de stijfheid van het SI gewricht te meten, deze werd in vitro getest. In deze studie werden vier gebalsemde vrouwelijke bekkens geëxciteerd door een bekkentriller. Een CDI scanner werd gebruikt om de trillingswijdte op de verschillende posities van het bekken weer te geven. Trillingen werden unilateraal aangebracht op de spina iliaca anterior superior en werden ontvangen door de CDI over de gehele ipsielaterale SI oppervlakte. Drie verschillende condities van stabiliteit werden gecreëerd in de SI gewrichten; geen interventie, geschroefd en met doorgesneden ligamenten. Onderzoeksresultaten werden gekwantificeerd door het nemen van de minimum drempelwaarden van de beenderen. Het relatieve verschil van trillingintensiteit tussen het ipsielaterale ilium en het sacrum in elke conditie van stabiliteit werd geaccepteerd als het stijfheidsniveau van het SI gewricht. Statistieken toonden hoge herhaalbaarheid aan alsmede belangrijke verschillen tussen de condities van stabiliteit. Dynamische testmethoden gebaseerd op het gebruik van trillingen brachten zichtbare en kwantificeerbare intra- en inter- individuele verschillen tussen stijfheden van het SI gewricht aan het licht. Deze nieuwe methode

is objectief en herhaalbaar. Toekomstige in vivo toepassing is veelbelovend daar er geen technische of veiligheidsrestricties zijn.

Studie drie betrof primair de peripartum bekken- en de lage rug pijn welke een veelvoorkomende klacht is bij vrouwen. De etiologische relatie tussen pijn en bekkenstabiliteit is in eerdere studies aangetoond, maar op dit moment is er geen objectieve klinische testmethode om de bekkenstabiliteit te evalueren. In deze studie werd een dynamische meetmethode, met gebruik van sonoelasticiteit, gebruikt voor het vaststellen van de stijfheid van het SI gewricht en in vivo getest op 14 gezonde vrouwelijke proefpersonen. Op de personen in rugligging werden trillingen unilateraal toegepast op de spina iliaca anterior. De trillingen werden geregistreerd door een CDI transducer op het ipsielaterale SI gewricht. Daar het niveau van de drempel van het apparaat rechtstreeks in verband staat met de kracht van de trillingen werd de intensiteit van de trillingen (sonoelasticiteit) op het sacrum en ilium indirect gemeten in drempeleenheden. De verschillen tussen de drempelwaarden werden geaccepteerd als het krachtverlies van de trillingen door het SI gewricht. De One-way Analysis of Variance-test and T-test for Paired Samples werden toegepast op de metingsresultaten ($P<0.05$). Statistisch gezien vertonen de resultaten een bevredigende intra-individuele herhaalbaarheid en inter-individuele veranderlijkheid. Er was geen belangrijk verschil tussen de data verkregen van het linker SI gewricht en die van het rechter SI gewricht. Gezien de veelbelovende resultaten bij gezonde vrouwelijke proefpersonen zal deze methode in toekomstige studies met name worden gebruikt voor patiënten met peripartum bekken pijn (peripartum pelvic pain, PPPP).

In studie vier werden de verschillen tussen de stijfheden van SI gewrichten tussen PPPP patiënten en gezonde proefpersonen vergeleken met behulp van Doppler Imaging of Vibrations (DIV). PPPP is een conditie gerelateerd aan de pathologie van SI gewrichten na uitsluiting van andere oorzaken, zoals lumbale discus hernia of spondylolisthesis. Alhoewel fysiek onderzoek en radiologisch onderzoek (X-rays, computerized tomography en MRI) informatie geven over de pathologie van de lumbale wervelkolom en het bekken, dragen deze technieken niet bij tot het inzicht in abnormale biomechanica. In eerdere studies introduceerden wij een nieuwe techniek om stijfheid van het SI gewricht vast te stellen met gebruik van CDI en trillingen, genaamd DIV. Deze techniek werd eerst getest op een plastic model, gebalsemde menselijke bekkens en gezonde proefpersonen. In deze studie werden de metingen uitgevoerd op een groep van PPPP patiënten (n=56) en een groep gezonde proefpersonen (n=45). Een triller, Derritron VP3, en een Colour Doppler Imaging apparaat (Philips Quantum ADI) werden gebruikt. Een protocol vergelijkbaar met dat welke eerder werd gepubliceerd door onze groep werd gevolgd. Het verschil in stijfheid van het SI gewricht tussen de groep patiënten en de groep proefpersonen werd statistisch getoetst door middel van de Wilcoxon's two sample test, de chi-square test en Student's t-tests. Zowel de patiënten als de gezonde proefpersonen vertoonden hypo- alsmede hyper- mobiliteit, maar er was geen belangrijk verschil tussen deze groepen. Er was echter wel een belangrijk verschil tussen deze groepen met betrekking tot het gemiddelde verschil in stijfheid van het linker en rechter SI gewricht. Asymmetrische stijfheid van de SI gewrichten leek meer rechtstreeks gerelateerd aan PPPP dan het niveau van stijfheid van een enkel SI gewricht.

Studie vijf betrof de beeldweergave van de bewegingen van pezen en spieren van de hand en pols van gezonde proefpersonen met CDI. De meeste beeldweergavetechnieken welke worden gebruikt om het bewegingsapparaat te onderzoeken, zoals X-ray, CT scan en MRI, produceren voornamelijk statische beelden. CDI biedt de mogelijkheid om dynamische karakteristieken van pezen en spieren te bestuderen. Het doel van deze studie was om de mogelijke toepassing van CDI in de beeldweergave van het bewegingsapparaat te onderzoeken. Een Philips Quantum Doppler werd gebruikt om de functionele anatomie van het bewegingsapparaat van gezonde proefpersonen te bestuderen. Deze studie resulteerde in een preliminair rapport van het Colour Doppler onderzoek en de beeldweergave van pols en hand. De resultaten geven aan dat CDI veelbelovend is voor de dynamische beeldweergave van de pezen en spieren in het bewegingsapparaat.

Het doel van studie zes was het onderzoeken van de unilaterale flexor en extensor aandoeningen van de hand en de verkregen beelden te vergelijken met die van de gezonde zijde. Een Philips Quantum Doppler werd gebruikt voor het onderzoek van patiënten met peesverklevingen en trigger finger. De

normale en pathologische pezen werden gescand op drie anatomische niveaus; pols, metacarpale en proximale phalanx. De resultaten verkregen door het vergelijken van pathologische beelden met normale, geven aan dat CDI veelbelovend is voor de dynamische beeldvorming en mogelijkerwijs, in de toekomst, gebruikt kan worden voor het kwantificeren van de functie van de pezen en spieren in het bewegingsapparaat en het definiëren van hun aandoeningen.

Het doel van de zevende studie was om CDI te gebruiken voor metingen van de maximale en gemiddelde peessnelheid. Eerdervermelde studies hebben uitgewezen dat CDI gebruikt kan worden voor de beeldweergave van pezen in de hand en pols. Alhoewel andere technieken beschikbaar zijn voor de beeldvorming van het bewegingsapparaat, is het in vivo meten van de snelheid van de peesverplaatsing niet mogelijk. De flexor pollicis longus (FPL) pees van 16 gezonde proefpersonen werd bilateraal gemeten op twee niveaus (pols en thenar). Een spalk werd van de vingers langs de proximale onderarm aangebracht. De duim werd bij de eerste phalanx gefixeerd aan de spalk opdat slechts het buigen van het interphalangeale IP gewricht mogelijk zou zijn. Pulsed CDI werd gebruikt voor de metingen. De maximale en gemiddelde snelheid van de FPL pees werd gemeten op een spectrale weergave stand gedurende continue actieve bewegingen van de vinger. Tenminste tien Doppler pieken (cm/sec) achter elkaar werden geregistreerd bij elke proef. De metingen werden drie maal herhaald. De Paired t-test werd toegepast en correlatie coëfficiënten werden berekend tussen niveaus op dezelfde zijde en de andere zijde. Er werden geen belangrijke verschillen gevonden tussen de twee niveaus van dezelfde hand en deze van de andere hand. Zoals verwacht toonde de data variaties aan in de inter-individuele peessnelheid. De snelheid van de verplaatsing van de FPL pees kan worden gemeten en goed weergegeven met CDI. Naar verwachting zullen de snelheidsmetingen in de toekomst kunnen worden gebruikt voor het vaststellen van andere pezen met aandoeningen.

Het doel van de achtste studie was het vaststellen van de toepassing van peesverplaatsingsmetingen met CDI bij gezonde proefpersonen en patiënten met hand aandoeningen, en het presenteren van de preliminaire resultaten. Vier gezonde proefpersonen en acht patiënten namen deel aan de studie. Patiënten met verschillende hand afwijkingen werden geselecteerd in onze hand polikliniek van de afdeling revalidatie. Een Philips ADI (Angio Dynagraph I) met software versie 2.5 gepulste multichannel CDI scanner werd gebruikt. Patiënten werden slechts één maal bilateraal gemeten maar hetzelfde metingsprotocol werd drie keer herhaald voor gezonde personen. Variance analyse voor metingen van buigingen gaven een belangrijk verschil aan tussen de personen ($p<0.01$), geen veel betekenend verschil tussen de beide zijden ($p=0.70$) en een belangrijk verschil tussen de pezen ($p=0.02$). Dezelfde testen voor metingen van strekkingen gaven aan dat er een groot verschil was tussen de personen ($p<0.01$), geen veel betekenend verschil tussen de zijden ($p=0.85$) en ook geen veel betekenend verschil tussen de pezen ($p=0.10$). De Student's t-test gaf een belangrijk verschil te zien tussen de gezonde zijde en de zijde met beperkingen ($p<0.01$). De belangrijkste conclusies van deze studie zijn: pees verplaatsingen uitgedrukt in millimeters kunnen non invasief gemeten worden met CDI; grote inter en intra individuele vinger buiging/strekkingspatronen bestaan bij gezonde personen; in deze preliminaire studie werden belangrijke links-rechts verschillen gevonden bij patiënten welke overeenstemden met de klinische onderzoeksresultaten.

Het doel van studie negen was het vergelijken van Doppler metingen met waarden van een verplaatsingsmeter welke rechtstreeks verbonden was met een riem. In geval van verschillen, was het doel de nodige filters en correctie constanten te ontwikkelen alsmede te testen om de precisie te verbeteren. Het model bestaat uit een rubber riem bewegend in een met vloeistof gevulde container. Buigen en strekken van de pees werden gesimuleerd door het met de hand bewegen van de riem. De riem was vastgemaakt aan de digitale verplaatsingsmeter voor kalibratie van geregistreerde bewegingen. Een multi-channel gepulste Colour Doppler Imaging apparaat (CDI) werd in het experiment gebruikt. Doppler signalen werden eerst overgebracht naar een envelop detector en dan naar een PC waar Doppler snelheidssignalen waren geïntegreerd tot waarden van verplaatsing van de riem. Gemiddelde en standaard afwijkingen van de verplaatsingswaarden in deze sets data werden geschat op $2.3\pm0.3$ cm/sec en $2.1\pm0.3$ cm/sec voor buigen en strekken respectievelijk. Deze gemiddelde waarden zijn tussen 20%

en 15% respectievelijk onder de uitgevoerde beweging van 2.7 cm. Lineaire regressie analyse leverde helling op van 0.012 cm/sec en 0.013 cm/sec respectievelijk, alleen een geringe afhankelijkheid van de gemeten verplaatsing van de piek snelheid indicerend.

Het doel van studie tien was het vaststellen van de toepassing van de pees verplaatsing mede door middel van CDI bij mensen, dit vormt een medium voor metingen welke de in vivo condities dichter benaderen dan een fantoom model, en het vaststellen van de correlatie tussen de met Doppler gemeten waarden en degene verkregen met behulp van de verplaatsingsmeter. Experimenten werden uitgevoerd op een gebalsemde arm van een mens. Flexor pollicis longus (FPL), flexor digitorum superficialis drie (FDS3) en flexor digitorum profundus twee en drie (FDP2 en FDP3) spieren werden apart bevestigd aan een gewicht van 10 N met een stalen kabel lopend over een katrol. Dit gewicht bewoog het telescopisch einde van een digitale verplaatsing meter op en neer tijdens de handmatige strekking en buiging van de vingers van het specimen. Verplaatsingsmetingen met Doppler werden uitgevoerd met een gepulste multi-channel CDI scanner. Doppler signalen werden overgebracht naar een envelop detector en dan naar een PC waar Doppler verplaatsingssignalen werden geïntegreerd in waarden van peessnelheid. Statistisch waren de Doppler metingen 3% lager, $p<0.00$ voor zowel buiging als strekking. CDI metingen van de vinger pees strekking correleerden in hoge mate met een mechanische micro verplaatsingsmeter, deze laatste word geacht een gouden standaard te vormen. Pees verplaatsingsmetingen met CDI op een menselijke hand en arm geven accurate resultaten. Kalibratie waarden welke worden gebruikt in de software om CDI signalen te converteren in pees verplaatsingswaarden in vivo geven een goede correlatie met de gouden standaard zoals gebruikt in de kadaver experimenten.

De hiervoor beschreven studies hebben op de eerste plaats aangetoond dat CDI klinisch zinvolle mogelijkheden biedt op het gebied van het bewegingsapparaat. Technisch gesproken zijn er aantal eigenschappen van CDI van dit doel onverwacht gunstig gebleken. Zo blijkt pees en spier excursies goed te kunnen worden gekwantificeerd. Tijdens validering studies bij gezonde personen en patiënten zijn ook nieuwe klinische inzichten verkregen, in het bijzondere met betrekking tot lage rug en bekken klachten. Dit was mogelijk dankzij de introductie van een nieuwe onderzoek techniek, Doppler Imaging of Vibrations (DIV).

## Curriculum Vitae

Muzaffer Buyruk was born in Istanbul, Turkey, in 1963. He received the Doctor of Medicine degree in 1987 and Specialist in Anatomy degree in 1991, both from Istanbul University Faculty of Medicine. After working as a member of teaching staff in the same faculty, he had started with his Ph.D. study in the Department of Rehabilitation, Erasmus University, Rotterdam, The Netherlands since 1992. His research interests include the Doppler imaging of musculoskeletal system, physical properties of pelvic girdle, biomechanics of lumbar spine and pelvis, and the relation of low back pain to sacroiliac joints. At the present he is working as resident in the same department.

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