Unsupervised Geometrical Modelling of the Scoliotic Spine from Radiographs

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Thesis Proposal

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Abstract

Scoliosis is a three-dimensional (3D) deformation of the spine that requires 3D examinations for correct evaluations. However, conventional 3D imaging techniques are not adequate for scoliosis evaluation. Therefore, physicians usually prescribe two planar radiographs for assessing this condition.

For dealing with this issue, some methods have been proposed for constructing 3D models of the spine using radiographs. The first problem when dealing with radiographs is determining the real dimensions of the structures that they capture. Currently, this is solved using radiopaque calibration objects. Unfortunately, these objects alter images content and require considerable changes in radiological environments and protocols. After calibrating radiographs another problem arises: how to capture vertebrae shape, position and orientation. Most of the methods require a set of points to be handmarked by an user on every vertebrae and on both radiographs, which is time-consuming and resource-consuming.

We propose developing an unsupervised method for constructing personalised geometrical models of scoliotic spines using radiographs acquired in standard radiological environments. For achieving this goal we will first try to create a calibration method that does not rely on calibration objects, and where user intervention will be minimised. Segmentation algorithms will be used for reducing or eliminating handmarked points. Additionally, we will try to reconstruct vertebrae using Deformable Shape Models (DSM). DSMs will be assisted by the output of the previous segmentation, which will provide a good initial solution and thus will increase the probability of modelling vertebrae correctly.

With this thesis we hope to contribute with a method that enables better scoliosis evaluations, and that may be used in standard radiological environments without special calibration equipment and without requiring considerable user intervention.
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Chapter 1

Introduction

1.1 Scoliosis and its clinical evaluation

According to the Scoliosis Research Society, scoliosis is defined as a lateral deviation of the normal vertical line of the spine which, when measured by X-ray, is greater than ten degrees (figure 1.1) (SRS, 2000). In fact, this traditional definition is limited since the deformity occurs in varying degrees in all three planes: back-front, side-to-side, and top-to-bottom. Additionally, vertebrae are rotated along the lateral curvature (axial rotation), and other structures are also affected, such as the rib cage. For a better understanding of the content of this proposal, we have included a representation of the basic anatomy of the spine (figure 1.2).

![Figure 1.1: Representation of a normal and a scoliotic spine.](image)

For evaluating a three-dimensional (3D) deformation of the spine, such as scoliosis, one would expect that 3D examinations, such as Computed Tomogra-
PHY (CT) and Magnetic Resonance Imaging (MRI), would be the elected ones. However, these methods have several disadvantages that make them unsuitable to capture the spine geometry. For starting, both methods oblige the patient to be lied down, which alters the spine posture. When evaluating scoliosis the patient should be in erect position (Cassar-Pullicino and Eisenstein, 2002; Fairbank, 2002; Greenspan, 2004). Additionally, both methods are expensive, especially MRI. CT scans are sensitive to bone tissue (Rogers, 1998), but it is impracticable to use CT for scoliosis evaluation due to the high doses of radiation needed to cover the entire spine. CT scans are only used to exam a limited set of vertebrae when a highly detailed reconstruction is needed (e.g. vertebrae fracture detection) (Greenspan, 2004; Rogers, 1998). As for MRI, it is a non-invasive technique that enables to capture the patients anatomy in several planes and make 3D reconstructions of internal structures (Greenspan, 2004). Unfortunately, this powerful imaging technique is unable to capture cortical bone* (Rogers, 1998) and therefore is not adequate for capturing the geometry of bone structures with detail. However, MRI is an important tool for assessing neurological complications that may be originated by the irregular curvature of the spine, and is often used as a complementary diagnosis examination (Cassar-Pullicino and Eisenstein, 2002; Greenspan, 2004; Fairbank, 2002).

For the above reasons, planar radiography is still the standard technique for evaluating scoliosis. Radiography is much less invasive that CT scans and is

* Cortical bone is the outer layer of a bone, which defines its surface.
1.2. Extracting geometrical information from radiographs

One of the main problems when extracting geometrical information from radiographs is mapping radiographs coordinates into real-world coordinates. This happens because of several factors. For starting, radiographs are 2D projections and therefore have no depth information. For solving this problem one may use more than one projection. The most common approach is to use the two orthogonal projections that usually are requested by physicians: one lateral and one frontal (AP or PA). Using the 2D coordinates of points that are visible in both

sensitive to bone structures, which enables to capture the entire spine. The standard (and minimum) radiographs that are prescribed are one lateral radiography (left or right) and one frontal (Anterior-Posterior (AP) or Posterio-Anterior (PA)) (Cassar-Pullicino and Eisenstein, 2002; Fairbank, 2002; Greenspan, 2004). These radiographs enable to evaluate the spine curvature in two orthogonal planes but still do not provide complete 3D information of the pathology. Some parameters, such as the maximum degree of the spine curvature or the axial rotation of each vertebrae may be difficult to assess.

For dealing with these problems some methods have been proposed for reconstruction the spine structure in 3D using the two radiographs that usually are prescribed (figure 1.3). These methods try to give physicians a 3D visualisation and 3D measurements without subjecting the patient to further examinations and to further radiation. In the following section we will introduce the reader to the main challenges related with this kind of techniques.

Figure 1.3: Construction of a personalised geometrical model of the spine using the two radiographs usually prescribed by physicians.

1.2 Extracting geometrical information from radiographs

One of the main problems when extracting geometrical information from radiographs is mapping radiographs coordinates into real-world coordinates. This happens because of several factors. For starting, radiographs are 2D projections and therefore have no depth information. For solving this problem one may use more than one projection. The most common approach is to use the two orthogonal projections that usually are requested by physicians: one lateral and one frontal (AP or PA). Using the 2D coordinates of points that are visible in both

projections it is possible to calculate their 3D coordinates. However, these coordinates do not correspond to real-world coordinates because radiographs suffer from a scaling effect that must be corrected. X-ray machines use a static x-ray source that emits diffuse radiation in order to capture the entire target in one shot only. The radiation passes through the patient body and is then captured in a radiographic film (figure 1.6). Unfortunately, this method causes changes in the size of the radiographed objects that depend on the objects position relatively to the position of the x-ray source and to the position of the radiographic film. In figures 1.4 and 1.5 it is possible to observe the scaling effect of using a diffuse x-ray source. In figure 1.5 in particular we can see that this scaling is not linear. In spine radiographs the scaling effect has a great impact because the target body area is very large and because the distance between vertebrae and the film is not constant due to the curvatures of the spine, either natural or pathological (figure 1.6). Additionally, the position of the x-ray source varies from radiography to radiography according to the patients anatomy, and therefore it is difficult to have a general estimation of the scaling factor.

Figure 1.4: Effect of the horizontal distance between the object being radiographed and the film plane. Higher distances from the film lead to higher scaling of the object when projected to the radiographic film.

For tackling the above problems, Calibration methods were already developed. These methods, which will be covered in detail in the next chapter, usually use calibration objects with known dimensions that appear in radiographs and enable to calculate the geometrical transformation that should be applied to x-ray coordinates in order to convert them into real-world coordinates. However, the most used calibration apparatus for calibrating spine radiographs need a considerable adaptation of the radiological environment that is neither practical nor affordable. Such adaptation is based on using a calibration apparatus with sufficient dimensions to surround the patient (Dansereau and Stokes, 1988; Dumas et al., 2003).
1.2. EXTRACTING GEOMETRICAL INFORMATION FROM RADIOGRAPHS

Figure 1.5: Effect of the displacement of the object being radiographed relatively to the film centre. Objects far from the centre of the x-ray film suffer a non-linear scaling effect when projected to the radiographic film.

Figure 1.6: Illustration of a spine being radiographed. Spine radiographs cover a very large area, and distances between vertebrae and the film are not constant, which worst distortions caused by the scaling effect.

Currently, most of the methods developed for construction geometrical models of the spine implemented in radiological environments with these features.

Other systems use small calibration objects and have the disadvantage of altering the radiological protocol and introducing objects in radiographs that superimpose anatomical structures (Cheriet et al., 1999b; Kadoury et al., 2007b). Finally, other approaches try not to use any calibration object, but the results were unsatisfactory because the scaling error was significant (Cheriet et al., 1999a; Kadoury et al., 2007a). Either way, and as far as we know, all algorithms for
constructing geometrical models of the spine for radiological environments without calibration objects or with small calibration objects require considerable user intervention for defining landmarks in every vertebrae for each radiograph, which makes of them very tedious, delayed, and resource consuming.

In terms of geometrical reconstruction of the spine, several methods are available that also will be covered in detail in the next chapter. The most accurate methods are still very manual. The oldest from Mitulescu et al. (2002) require a trained technician to spend 2 to 4 hours per patient for reconstructing the entire spine and pelvis. Another more recent method reduces user intervention to about 16 minutes per patient and achieves slightly better reconstruction results Pomero et al. (2004), although vertebrae orientation is not so accurately calculated. As far as we know, only one automatic method was proposed (Benameur et al., 2005) but it was only applied to the lower half of the spine, which is the clearer part of spine radiographs.

Summarising, currently there is no method capable of performing accurate reconstructions of the entire spine (or at least the thoracic and lumbar spine) automatically, or with very limited user intervention. In terms of calibration, there is no method for accurately calibrating spine radiographs without using calibration objects. Moreover, when using small calibration objects, the current methods require considerable user intervention. Our goal is trying to close some of this gaps by developing algorithms and methods that enable to automatically perform geometrical reconstructions of the spine using radiographs while minimising the need for calibration objects. By accomplishing this goal we expect to make this technology available to most of the health institutions without the need of acquiring expensive equipment nor allocating significant human resources.

1.3 Structure of the document

The remainder of this thesis proposal is outlined as follow:

*Chapter 2* reviews the main literature on spine radiography calibration, and on construction of geometrical models of the spine from radiographs.

*Chapter 3* briefly states this thesis by defining its goal and emphasising its key issues.

*Chapter 4* proposes methods for accomplishing this thesis goal, and procedures for validating these methods.

*Chapter 5* describes the work plan for this thesis project, defining the principal tasks and their deadlines.

*Chapter 6* closes this proposal with a summary of the main implications and expected contributions of this thesis.
Chapter 2

State of The Art

In the previous chapter we have explained the motives that help us choosing radiographs as input for constructing geometrical models of the spine, and the main difficulties that this task presents. In this chapter we will review the main literature concerned with solving these problems. We will start by addressing spine radiography calibration and then the construction of geometrical models from radiographs. During the reading of this chapter, the reader is invited to consult table 2.4 (on page 33) that summarises the main methods reviewed in this chapter.

2.1 Calibration of spine radiographs

In section 1.2, we have described the problem of mapping radiographs coordinates into real-world coordinates, which is particularly affected by a non-linear scaling effect. In this section we will present the main approaches used to solve this problem.

Radiography calibration is a very similar problem to camera calibration from computer vision. The goal is to find the calibration matrix that is able to map image coordinates into world coordinates. The calibration matrix is composed by two set of parameters: intrinsic and extrinsic. Intrinsic parameters are related to the camera properties, and extrinsic parameters are related to the camera position. By determining these parameters one is able to convert any 2D point (present in two or more projections) in a 3D point of the real-world.

Several techniques from the computer vision community may be used to try solving the radiography calibration problem with few adjustments. In most of these techniques a calibration object with known dimensions is required for calculating the calibration matrix. From all the reviewed literature, the most used approach for spine radiography is using a large calibration object with sufficient
dimensions to surround a person. This approach yields accurate results but has several inconveniences, such as costs and portability (among others). For dealing with these problems, in the last years there have been efforts for reducing the size of the calibration objects or even eliminate them at all. In the next subsections we will review the most used approaches and the ones that have emerged in the last years.

2.1.1 Large calibration apparatus

Currently, the most used approach for calibrating radiographs for constructing geometrical models of the spine is using a large calibration object that is large enough to surround a person’s body (Aubin et al., 1997; Mitton et al., 2000; Bras et al., 2003; Benameur et al., 2003, 2005; Pomero et al., 2004). The first method which fits in this class was proposed by Dansereau and Stokes (1988). The authors proposed using a very well known technique in computer vision: Direct Linear Transformation (DLT) (Abdel-Aziz and Karara, 1971). Basically, the DLT algorithm is a least-squares optimisation procedure that calculates the most probable intersection location between two (or more) rays going from two (or more) points from two (or more) images to the camera. When we have corresponding points marked in two or more images, DLT tries to find the minimum distance between the rays (ideally it would be zero), and doing so it calculates the 3D coordinates of the point projected on those images. In order for DLT to be used, it first needs to be calibrated using a set of control points (usually a minimum of 16 points) with known coordinates. Therefore, there is a need for a calibration object that should be fixed in the radiologic environment with a set of radio-opaque pellets. Unfortunately, DLT presents significant extrapolation errors. Thus, for accurately estimating 3D coordinates, Dansereau and Stokes proposed a calibration object that was built sufficiently large to position any anatomical structured to be reconstructed inside its limits (Dansereau and Stokes, 1988).

In figure 2.1 it is possible to observe the actual form of the calibration apparatus. This apparatus includes a rotatory platform for minimising patient movements between radiography acquisitions, and for keeping an approximately constant distance between the x-ray source and the spine (figure 2.2).

The best results so far using this calibration device with state of the art algorithms for in vivo spine reconstructions were achieved by Mitulescu et al. (2002) that measured an error of 1.5mm mean point to surface distance when compared with TC scans reconstructions with accuracy of 1.1mm. Despite of the good results achieved with this calibration technique, it has some drawbacks that are worth to mention. For starting, it requires considerable changes in regular radiographic environments that may not be possible or affordable. Additionally, the calibration marks superimpose anatomical structures. Besides these problems, Cheriet et al. (1999a) documented that the physical characteristics of the
2.1. CALIBRATION OF SPINE RADIOGRAPHS

Figure 2.1: Large calibration apparatus proposed by Dansereau and Stokes (1988).

Figure 2.2: Diagram of the calibration apparatus proposed by Dansereau and Stokes (1988).

calibration object makes some subjects to be fear or feel uncomfortable, requires some patients to kneel, and are not adequate for patients lying down (e.g. during surgery) and neither for patients that are not able to stand up (e.g. patients in wheelchairs).

For trying to tackle some of the above problems, Dumas et al. (2003) developed a new calibration device and a calibration method that was not based on DLT. Instead, the authors developed a simplified geometrical model of the ra-
ological environment and made some assumptions that enabled to get a set of calibration equations. The main assumptions are: (i) the image reference plane must be parallel to the global reference frame in both radiographs (frontal and lateral), and (ii) the x-ray source must remain in the same position between radiographs. Having these assumptions in mind and the proposed calibration equations, the authors developed the calibration device that is represented in figure 2.3. As one may see, now the calibration marks rotate with the subject. The authors managed to escape from the DLT algorithm and therefore were able to developed a system where the patient does not need to be fully enclosed by the calibration object. This system is therefore more patient-friendly and has the additional advante of reducing the number of marks that superimpose bone structures. For evaluating the system accuracy, this method was compared with the previous. In particular, the authors used the same dried vertebrae and the same reconstruction algorithm presented in Mitulescu et al. (2001), which validated the previous method using dried lumbar vertebrae. The results were very similar: the new system was able to achieve a mean point to surface distance of 1.2mm (against 1.1mm), RMS of 1.6 (against 1.4mm) and maximum error of 6.4mm (against 7.8mm) (Dumas et al., 2003).

Figure 2.3: Large calibration apparatus proposed by Dumas et al. (2003).

2.1.2 Avoiding calibration objects

As we have seen in the previous section, calibration objects have several disadvantages, especially if they have large dimensions. In order to avoid using calibration apparatus, Cheriet et al. experimented calibrating radiographs without using any
apparatus. The input of the proposed method was a set of stereo-corresponding points (points that are visible in both images) that were manually marked over specific regions of every vertebra, and an estimation of the calibration parameters. These parameters were then optimised in order to minimise the mean square distance between the observed and analytical projections of the marked points. The authors experimented this method with angiography (Cheriet and Meunier, 1999) and spine radiography (Cheriet et al., 1999a). When calibrating angiography images, the authors had access to a very good estimation of some of the calibration parameters because the angiography machine gives information about the x-ray source position and the exact angle between the two acquisitions. This good estimation enabled to reduce the search space enough to find accurate calibration parameters (Cheriet and Meunier, 1999). However, when applying the same method to spine radiography, the authors did not have access to the x-ray source position, and neither to the geometric transformations between the two acquisitions. For filling this gap, the authors collected the calibration parameters of several radiographs taken with a large calibration object in order to have an average solution and a standard deviation. Using this information to choose an initial guess for the optimisation algorithm and to narrow the search space was proved to be insufficient for finding a good solution of the calibration parameters (Cheriet et al., 1999a). Having failed to calibrate spine radiographs, the authors attempt to find out the minimum absolute information that a calibration object should provide. For this task they used the calibration object presented in the previous section proposed by Dansereau and Stokes (1988). The conclusion was that for accurate results the algorithm should have access to the absolute coordinates of three coplanar points and to the absolute measure of four distances.

Using the same method, Kadoury et al. (2007a) tried to perform geometric spine reconstructions without using any calibration object. The results were that angular measures of the spine were achieved with small errors, but the same did not happen to absolute mesures, such as the spinal length, which had an error of $14.19 \pm 8.00$ mm. This shows that the method is not able of handling the scaling effect, because scaled objects present the same angular measures but the absolute mesures are obviously different.

Not using calibration objects has several advantageous, such as low costs, minimum adaptation of the radiological environment, minimum constrains for the patient position (e.g. the patient may be seated), and radiography images remain unaltered. To our knowledge, no method as proved to be accurate in calibrating spine radiographs without using calibration objects. Moreover, the methods described here require considerable user intervention.
### 2.1.3 Small calibration objects

Having failed to develop a method that does not rely on calibration objects, a small calibration object was constructed (figures 2.4 and 2.5) in order to obtain accurate calibrations during surgery Cheriet et al. (1999b). The calibration algorithm was the same suggested in Cheriet et al. (1999a). Until this moment, we were not able to get the article reporting experiments using this calibration object, but according to Novosad et al. (2004) the RMS reconstruction error of the spine using this calibration method is 2.5mm. This error is higher than the RMS reconstruction errors achieved using large calibration apparatus, which justifies the preference for large calibration objects when they are available. However, this error is still admissible for several clinical applications, especially when it is not possible to have patients standing up, like in surgery.

Source: adapted from Novosad et al. (2004)

**Figure 2.4:** Setup of the radiological environment for surgery when using the small calibration object proposed by Cheriet et al. (1999b).

Source: adapted from Novosad et al. (2004)

**Figure 2.5:** The small calibration object proposed by Cheriet et al. (1999b) has a set of 15 steel pellets of known positions.

Kadoury et al. (2007b) also developed a small calibration object, but now with the intent of being portable. The object has a set of four pellets with
known positions, and it is placed in a vest worn by the patient during both acquisitions (figures 2.6 and 2.7). The proposed method uses this object and the weak-perspective algorithm to extrapolate the geometrical parameters of the radiological environment. This technique does not provide accurate calibrations but enables to have a reasonable estimation of the parameters values. In order to improve calibration, the authors use the optimisation algorithm proposed by (Cheriet et al., 1999a), which needs a set of stereo-correspondent points. In this work, the authors used a method for constructing the geometrical model of the spine that needs a set of 6 points per vertebra to be manually identified in both images. The authors take advantage of these points and use a subset of them to feed the optimisation algorithm. Using this method the authors were able to improve the results of their previous work (Kadoury et al., 2007a) were they did not use any calibration object. In particular, they were able to reduce the error when calculating absolute measures, such as the spinal length that previously scored 14.19 ± 8.00 and now scored 2.05 ± 1.03. This indicates that this method is coping with the scaling effect problem. However, relative (angular) measurements obtained inferior results (although acceptable). Unfortunately, the authors did not publicate any data related to the accuracy of spine reconstruction and therefore a comparison with the other methods on this subject it is not possible.

**Source:** adapted from Kadoury et al. (2007b)

**Figure 2.6:** The small calibration object proposed by Kadoury et al. (2007b) has 4 pellets of known positions.

Small calibration objects have almost the same advantageous as not using calibration objects at all. However, using small calibration objects results in higher costs (the cost of the calibration object), adaptations to the radiological procedure, and more importantly, the objects overlap anatomical structures on radiographs (as it is possible to observe in figure 2.7). In terms of accuracy, using small calibration objects is obviously advantageous, although for best results one should use large calibration apparatus.

Besides the methods presented here, other solutions are used for general radiography. One standard technique is to use a small calibration sphere with known dimensions that should be aligned with the anatomical structure under exami-
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Figure 2.7: Radiographs showing the calibration object proposed by Kadoury et al. (2007b). All 4 pellets are visible in both radiographs.

nation. Using one sphere only it is possible to calibrate radiographs by applying the observed scaling effect. For instance, The (2006) was able to calibrate pelvis radiographs using one sphere that should be positioned between the patient legs and aligned with the pelvis. To our knowledge, this technique was never applied to spine radiography probably because of the large dimensions and irregularities of the spine that worsen the scaling effect.

2.1.4 Hybrid Methods

Recently, Cheriet et al. (2007) used a novel technique to calibrate spine radiographs. It consists in mixing two kind of calibration objects: one object with absolute positioning in the radiological environment, and a vest worn by the patient with a set of calibration pellets (figure 2.8). The first object offers an absolute reference plane that helps calculating calibration parameters. On the other side, the vest offers a set of points that follow the patient and enable to compensate undesired movements between acquisitions. Moreover, the vest offers accurate stereo-correspondent points that are more reliable and easy to localise than points hand-marked in specific anatomic regions of the patient. In this way, the calibration algorithm only uses positioning information provided by calibration objects (the fixed object and the vest), which is more accurate and easier to detect. Thus, the authors were able to use a method were calibration points were automatically detected (Kadoury and Cheriet, 2006). Currently, only preliminary results are available indicating that this method may be more robust to patient movements between radiographs comparing to large calibration apparatus, which only try to minimise patient movements using rotatory platforms. Additionally, relative measurements seem to be more accurate than the method proposed by Kadoury et al. (2007b). To our knowledge, neither absolute measurements nor results about the accuracy of geometrical models constructed with
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This calibration method were published.

Source: adapted from Cheriet et al. (2007)

Figure 2.8: The calibration vest proposed by Cheriet et al. (2007) has 16 pellets.

2.1.5 Summary

In this section we presented the state of the art of spine radiography calibration. We have seen that large calibration apparatus are still the gold-standard in terms of accuracy. Small calibration objects offer more flexibility and lower costs but are also less accurate. Nevertheless, small calibration objects offer sufficient accuracy for several clinical applications. Using calibration objects, either large or small, always have disadvantages, such as overlapping anatomical structures and costs. Thus, calibrating spine radiographs without using calibration objects would be an interesting approach, but presently no method was able to do this. More recently, a novel technique is emerging that combines calibration objects fixed to the radiological environment with a calibration vest worn by the patient. This new approach seems to be capable of compensating undesired patient movements between the two acquisitions, but no results concerning accuracy evaluation of absolute measurements are yet available.

2.2 Construction of geometrical models of the spine using radiographs

After calculating the calibration parameters for two (or more) radiographs of the same subject, one is able to calculate the real 3D coordinates for every point that is visible in both images (stereo-correspondent points). However, stereo-correspondent points are limited because some are occluded and others are difficult to match accurately in the pair of radiographs. Moreover, vertebrae have a complex geometry that difficult constructing geometrical models of the spine.

In the next subsections we will present the main approaches for constructing personalised geometrical models of the spine using radiographs. We will only
consider methods that try to construct morpho-realistic models and, therefore, older methods that create vertebrae models with a small number of points will not be reviewed.

2.2.1 Atlas deformation using handmarked landmarks

The methods of this subsection need a considerable user intervention, more precisely, several points must be handmarked for every vertebra in both radiographs. They were the first methods created for solving the problem of constructing geometrical models of the spine from radiographs. Some of them are still very used in clinical environment.

In general, a generic model (atlas) of the spine is deformed in order to fit to the patients spine. Deformations are controlled by the handmarked points. The generic model usually is created recurring to dried vertebrae (using 3D scanners or CT) or cadavers (using CT). Figure 2.9 shows an example of a generic model of a vertebra with approximately 200 points, which may be adjusted in order to fit to patients vertebrae.

\[ \text{Source: adapted from Mitulescu et al. (2002)} \]

![Figure 2.9: A generic model of a lumbar vertebra that is used as reference for constructing personalised lumbar vertebrae.](image)

Stereo-Correspondent Points and Kriging

The first method for constructing morpho-realistic models of the spine using radiographs was proposed by Aubin et al. (1997). This method needs 6 landmarks per vertebrae per radiograph that must be handmarked by a technician. These stereo-correspondent landmarks are 6 anatomical regions visible in both radiographs (figure 2.10). Using the DLT algorithm Abdel-Aziz and Karara (1971), the 2D coordinates of the stereo-correspondent points are converted into 3D coordinates. Thus, six real-world coordinates are known for each vertebra. For building morpho-realistic vertebrae models the authors used generic models of
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vertebrae (acquired from CT scans) that were transformed using dual kriging\(^*\) (Trochu, 1993) in order to fit the six landmarks as well as possible. This is done for every vertebrae.

Source: adapted from Mitulescu et al. (2002)

![Figure 2.10: Stereo-corresponding anatomical landmarks: the points marked in one view have a correspondent point in the other view and therefore the number of points is the same in both views.](image)

For validation proposes the authors used a cadaveric spine and compared a 3D reconstruction using a coordinate measuring machine (error about 0.1mm) with a 3D reconstruction using the proposed method with three radiographs (PA, PA 20° and Lateral). Point-to-surface errors were of 2.6 ± 2.4mm (mean ± standard deviation). It was also studied the effect of increasing the number of landmarks on the method accuracy. For testing this, experiments where done with 21 landmarks per vertebrae per radiography. The results were better but because of the exaggerated time required for handmarking all the points and because some of the points were sometimes difficult to see this idea was abandoned. The authors should have tested varying the number of landmarks in order to determine a good compromise between the time spent and the achieved accuracy, however such study was not performed.

Adding Non Stereo-Corresponding Points to SCP and Kriging

Using stereo-correspondent points with kriging is not enough for obtaining accurate reconstructions of the human spine. Adding more stereo-correspondent points besides the six points proposed by Aubin et al. (1997) may lead to more accurate reconstructions, however there are several important points of vertebrae anatomy that are only visible in one of the radiographs (when considering the standard frontal and lateral radiographs). Having this in mind, Mitton et al. (2000) proposed enhancing the previously describe method with non stereo-corresponding points (NSCP – points that are marked in only one of the radiographs). An example of non-stereo corresponding points is shown in figure 2.11, although in this work the authors tested their technique in cervical vertebrae (Mitton et al., 2000).

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\(^*\)Kriging is a technique for interpolating and extrapolating values for unobserved locations using information of nearby locations.
The main issue with NSCPs is that it is not possible to determine the 3D coordinate by triangulation because these points appear in one radiography only and therefore do not have a correspondent point in the other radiography. Thus, there is no way to determine the exact 3D coordinates in the real-world for NSCPs. The only information available is that the 3D coordinate will be somewhere in the line joining the x-ray source to the marked point in the radiography (constraint line). To overcome this problem Mitton et al. (2000) proposed a method where a generic object (a vertebra model) is used for guessing the 3D coordinates of NSCPs. This object is composed by a set of points connected by springs with a given stiffness that try to keep the deformed object as close as possible to the original object (in order to maintain a generic topology). For finding a solution an optimisation problem is defined where the goal is to minimise the total deformations, and the constrains are given by the SCPs and by the constraint lines. The authors used a conjugate gradient algorithm to compute the solution for the minimisation problem.

The algorithm for constructing geometrical models of the spine now has the following steps:

1. Manual identification of the SCPs (6 per vertebra per radiography) and NSCPs (15-26 per vertebra);
2. Calibration;
3. Computation of the 3D coordinates of the SCPs using DLT;
4. Computation of the 3D coordinates of the NSCPs;
5. Krizing of a generic object (with approximately 200 points) using the 3D points previously computed.

This algorithm was first applied for reconstructing dried cervical vertebrae (Mitton et al., 2000), then for dried lumbar vertebrae (Mitulescu et al., 2001) and finally for thoracic and lumbar vertebrae in vivo (Mitulescu et al., 2002).
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Table 2.1: Comparison of reconstruction errors of dried lumbar vertebrae for SCP (stereo-corresponding points only), NSCP (SCP + non stereo-corresponding points), NSCP with generic models of high detail, and CT scans. It is also presented the number of landmarks needed to be handmarked per vertebra, the number of points of the geometrical model for each vertebra, the number of vertebrae tested, and the validation method.

<table>
<thead>
<tr>
<th></th>
<th>CT</th>
<th>SCP</th>
<th>NSCP</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mean error (mm)</td>
<td>0.8</td>
<td>1.4</td>
<td>1.1</td>
</tr>
<tr>
<td>2RMS error (mm)</td>
<td>2.2</td>
<td>3.6</td>
<td>2.8</td>
</tr>
<tr>
<td>Maximum error (mm)</td>
<td>3.9</td>
<td>15.8</td>
<td>7.8</td>
</tr>
<tr>
<td>Number of landmarks</td>
<td>—</td>
<td>6 * 2 = 12</td>
<td>6 * 2 + 19 = 31</td>
</tr>
<tr>
<td>Model detail (points)</td>
<td>—</td>
<td>178</td>
<td>—</td>
</tr>
<tr>
<td>Test set size</td>
<td>—</td>
<td>18</td>
<td>—</td>
</tr>
<tr>
<td>Validation</td>
<td>3D scanner (±0.2mm)</td>
<td>—</td>
<td>CT (±1mm)</td>
</tr>
</tbody>
</table>

Results from the last study achieved reconstructions errors of 1.5±2.0mm (mean ± RMS), which shows a significant improvement over the previous technique. Additionally, Mitulescu et al. (2001) compared reconstructions using SCP only, SCP and NSCP, and TC scans against direct measures with a 3D scanner. The results demonstrated a significant improvements of mean and maximum errors, and also placed reconstruction using radiographs a step closer to TC scans reconstructions (table 2.1).

Later, Bras et al. (2003) increased the detail of generic vertebrae models from ~200 points to 2000 points (per vertebra). The original generic model had been obtained by direct measurements using a 3D scanner over a large set of dried vertebra (~1000 vertebrae) and was updated using a CT scan (2.12). This improvement seems to reduce the reconstruction error. However the validation technique was inferior and the quality of the radiologic equipment was superior, which difficult comparing results. Nevertheless, the maximum reconstruction error was significantly decreased, regardless of the accuracy of the validation technique (table 2.1).

NSCPs enabled to improve the quality of the geometrical reconstruction of the spine but increased the number of points that must be handmarked. Consequently, the time technicians must spent to reconstruct a spine is also increased and varies between 2 to 4 hours, depending of the image quality and of the patient pathology. This is a factor that obviously difficult implementing this method in clinical environment especially because of the required human resources.
2.2.2 Using a priori statistical knowledge

The methods from the previous section have several problems: (i) they need considerable user intervention (ii) they only use information about the hand marked points and ignore the image content (which provides valuable information about vertebrae shape) and (iii) they do not use information about shape irregularities on pathological vertebrae. The following methods tackle some of these problems using statistical knowledge in two very distinct ways: the first uses statistical information to estimate most of the vertebrae characteristics ignoring the image content, and the second deforms generic objects in order to fit them to the image content.

Semi-Automated Method

Statistical methods require a large dataset of vertebrae models with healthy and scoliotic vertebrae in order to correctly infer vertebrae shape. For this reason, the Semi-Automated method (SA) proposed by Pomero et al. (2004) uses a dataset of 3202 vertebrae, where for each vertebra it is stored its geometrical model (∼200 points) and measurements of the vertebral body. The authors claim that the spine may be reconstructed using this dataset and knowing only 4 points per vertebrae per radiography (making a total of 8 points per vertebra). These points correspond to the 4 corners of every vertebrae, independently of its rotation (figure 2.13). Using these points it is possible to calculate the box (more precisely the hexahedron) that encloses every vertebral body. With this box the proposed algorithm first calculates vertebrae lateral and sagital angulation. Then, axial rotation is estimated using a statistical model that relates it with the spine curvature. At this point, the position and orientation of every vertebra is known. For estimating vertebrae shape, a set of descriptors is calculated from the enclosing
box of every vertebra and is submitted to a linear regression model (based on the vertebrae dataset) that determines a set of 21 points. Finally, once again kriging is used to deform a generic vertebra model of 2000 points using the 21 points previously determined.

In this technique, almost every vertebrae characteristic is estimated using statistical knowledge, ignoring the information concerning vertebra shape provided by radiographs. For evaluating the accuracy of the proposed method, the authors used the same radiography set that was used by Mitulescu et al. (2002) to evaluate the NSCP technique. The accuracy of the statistical method was slightly superior and achieved a significant decrease of the maximum error (table 2.2). However, in terms of axial rotation (which is estimated in this method) results were not so good. Table 2.3 shows the differences between the proposed method (SA) and NSCP, and the orientation errors for the NSCP technique (for the study of Mitulescu et al. (2002) there was no absolute reference because it was an in vivo study, and therefore we also present the results of an orientation accuracy evaluation of the NSCP technique). As it is possible to see, the proposed method presents significant standard deviations from the NSCP technique (which seems to be more accurate), especially for the axial rotation that is estimated in the SA method using a statistical model.
CHAPTER 2. STATE OF THE ART

Table 2.2: Comparison of reconstruction errors of 58 vertebrae of 14 patients for NSCP and SA (Semi-Automated method). It is also presented the number of landmarks needed to be handmarked per vertebra, and the user time required to do it.

<table>
<thead>
<tr>
<th></th>
<th>NSCP</th>
<th>SA</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mean error (mm)</td>
<td>1.5</td>
<td>1.4</td>
</tr>
<tr>
<td>2RMS error (mm)</td>
<td>4.0</td>
<td>3.6</td>
</tr>
<tr>
<td>Maximum error (mm)</td>
<td>19.7</td>
<td>15.8</td>
</tr>
<tr>
<td>Number of landmarks</td>
<td>6 × 2 + 19 = 31</td>
<td>4 × 2 = 8</td>
</tr>
<tr>
<td>User time required</td>
<td>2–4h</td>
<td>~ 20m</td>
</tr>
</tbody>
</table>


Table 2.3: Orientation difference between NSCP and SA, and NSCP orientation errors.

<table>
<thead>
<tr>
<th></th>
<th>SA vs NSCP</th>
<th>NSCP error</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Mean ± SD</td>
<td>Mean ± RMS</td>
</tr>
<tr>
<td>Lateral</td>
<td>0.2 ± 3.6</td>
<td>0.6 ± 0.8</td>
</tr>
<tr>
<td>Sagital</td>
<td>−1.1 ± 3.7</td>
<td>0.7 ± 1.0</td>
</tr>
<tr>
<td>Axial</td>
<td>−1.3 ± 6.4</td>
<td>1.4 ± 1.9</td>
</tr>
</tbody>
</table>

In terms of reliability, Gille et al. (2007) studied the variation of clinical parameters obtained from the geometrical reconstructions for (i) the same user and the same radiographs at different times (intra-reliability) and (ii) for different users using the same radiographs (inter-reliability). The authors concluded that the observed variations were inside the expected range when analysing mild scoliotic patients, and in fact, almost all of these measures achieved high correlations in intra- and inter-reliability analysis.

The method proposed by Pomero et al. has comparable accuracy to NSCP (although vertebrae orientation may need some improvements) and requires much less user intervention. The number of landmarks was very reduced, and these landmarks are easier to mark because they have good visibility and do not require the user to imagine the location of vertebrae regions that may be difficult to see (like in NSCP). In addition, automatically detecting the corners of vertebrae seems to be a much easier segmentation task. In fact, Deschenes and de Guise (2002) have proposed a segmentation algorithm which is used in Deschenes et al. (2003) for detecting these 4 points per vertebra per image based on a small set of points handmarked along the spine and two line-segments also marked by the user. Unfortunately, we have not got access to this last document, and to our knowledge no results have been published about the reconstruction accuracy when combining SA from Pomero et al. (2004) with the semi-automatic point detection from Deschenes et al. (2003). Moreover, in an recent anthropometric study of exterior and interior geometry of 64 adults performed by the same institution
(Bertrand et al., 2006), the more manual version of this method was used, which possibly indicates that the more automatic method achieves inferior results or that it was not conveniently evaluated.

In our opinion the presented method would benefit from directly using the radiographs content, while the presently achieved reconstruction would provide excellent initial solutions for an optimisation process.

**Statistical deformable models**

Benameur et al. proposed combining contour extraction from radiographs with *a priori* knowledge about vertebrae shape (Benameur et al., 2001, 2003). This knowledge was captured in a statistical deformable template (one template per vertebra) that integrates a set of admissible deformations related to pathological changes observed on a representative scoliotic vertebra population (figure 2.14). For deforming templates, the authors used the contours of each vertebra, which were extracted from radiographs using a popular edge-detection algorithm (Canny, 1986). Then, an optimisation problem was defined where an energy function (calculated using vertebrae countours and a projection of the deformed template) was minimised using a gradient descent algorithm.

![Figure 2.14: Visualisation of a mean shape for the T6 vertebrae (middle column) and two deformations (left and right columns) from the sagittal (top row) and coronal views (bottom row).](source: adapted from Benameur et al. (2003))

The process just described enables to deform vertebrae templates until they fit in patients vertebrae, but in order to avoid trapping in local minimum it needs a close initialisation. The authors used a technique that estimates the position of 6 anatomical landmarks per vertebra (thoracic and lumbar) based on 8 points marked in every radiograph (16 points for two radiographs). This technique allowed to have an initial estimation of vertebrae shape and position, but during
experiments it was verified that this estimation was not always enough for escaping local minimum. The authors solved this problem creating variations of the initial estimation and thus providing several initial solutions for the optimisation algorithm. However, the success of this approach seems to be very sensitive to the parameters used for obtaining the initial solutions.

Using the described method the authors achieved results comparable with the other methods. These results seem to be even better at a first glance, but when we carefully analysed them we observed that for lumbar vertebrae only one or two vertebrae were validated per level (two L1, one L2, two L3, zero L4, one L4). We believe that this is a very small number of vertebrae per level, especially when each level has a specific template. Nevertheless, the authors obtained mean error of $0.71 \pm 0.06$ and maximum of $4.91 \pm 0.15$ for lumbar vertebrae. Another odd observation is that the maximum error presented is not the absolute maximum but rather an average of the maximum errors of every vertebrae reconstruction, and therefore it is difficult to compare these results with the presented by the other authors. For thoracic vertebrae more specimen were available (reaching 10 vertebrae for the T8 level) and the mean error was $1.48 \pm 0.15$ and the maximum was $7.28 \pm 4.00$. According to the authors, higher errors in the reconstruction of thoracic vertebrae are related to overlapping bone structures (e.g. ribs) that produce other countours that distract the optimisation procedure. We suppose that was also for this reason that studies using this method were limited to vertebrae T6 to L5 (in the upper half of the thoracic vertebrae, T1 to T6, there are more overlapping structures). Even though, the proposed method seems to achieve good results for the selected range of vertebrae requiring only 16 points to be handmarked per patient.

Later, Benameur et al. (2005) proposed a more unsupervised method where only 2 points per radiograph are needed (making a total of 4 points per patient). In this approach the authors used a statistical template for registering the complete spine (figure 2.15) and consequently determining an initial solution (instead of using the previous algorithm that requires 8 points per radiograph). Then, a similar process to the one used in the previous approach (although with some variations) is applied for accurately finding every vertebra shape. This technique leads to two separate minimisation procedures solved with a exploration/selection stochastic algorithm. The accuracy was inferior when compared with the previous method using the same test set, but remains comparable with the methods from the previous section. Only the maximum errors achieved for thoracic vertebrae were inferior. It is important to mention that both methods used a database of 30 normal and 30 scoliotic vertebrae, which certainly does not capture the complete variety of deformations and therefore influences reconstruction results.
2.2. CONSTRUCTION OF GEOMETRICAL MODELS OF THE SPINE USING RADIOGRAPHS

31

Source: adapted from Benameur et al. (2005)

Figure 2.15: Deformable model of the whole spine (a) and cubic template representation of a vertebra (b).

2.2.3 Summary

In this section we presented the state of the art of constructing geometrical models of the spine using radiographs. We have seen that two distinct methods currently offer the best results: NSCP (Mitton et al., 2000) and SA (Pomero et al., 2004). The first requires considerable user intervention (2–4 hours per patient) for marking a large set of points (∼31 points per vertebra) where some of these points are present in both frontal and lateral radiographs, and other only appear in one of them. No image analysis is performed by this method that is based in deforming generic vertebrae using the handmarked control points. The second method, requires much less user intervention (∼20 minutes per patient) because only requires 8 points per vertebra (the corners of vertebrae in both images) that are easy to identify. Then, almost all of the vertebrae characteristics are inferred using statistical models, and once again no image analysis is done for approximating the generated models to the image content. The accuracy in terms of vertebrae shape of this method is slightly superior to NSCP, but in terms of vertebrae orientation the results are considerably worst. In an attempt of using the image content, Benameur et al. (2003) used statistical templates of generic vertebrae that were deformed in order to best fit vertebrae contours extracted from radiographs. Benameur et al. presented results comparable with the previous methods, but only for the lower half of the spine (where the images are clearer). The method only required 16 points per patient but the authors reported difficulties in achieving an initial geometrical model with sufficient quality for the optimisation process to succeed. Later, Benameur et al. proposed a new method
were this initial solution was obtained more automatically. It used a statistical template of the whole spine and required a user to manually identify 4 points per patient. These approach scored inferior results than the more supervised version and still only covered the lower half of the spine.
### Table 2.4: Synthesis of the studies of geometrical models reconstruction of the spine with the correspondent calibration method.

<table>
<thead>
<tr>
<th>Reference</th>
<th>Sample Description</th>
<th>Calibration Method</th>
<th>Validation Method</th>
<th>Calibration</th>
<th>User Input</th>
<th>Error (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Aubin et al. (1997)</td>
<td>SCP, Large D88</td>
<td>SCP</td>
<td>D88 (±0.1 mm)</td>
<td>D88</td>
<td>17 pt/v</td>
<td>Mean=2.6 SD=2.4</td>
</tr>
<tr>
<td>Cheriet et al. (1999)</td>
<td>SCP, Small C99</td>
<td>SCP</td>
<td>Unknown</td>
<td>18 pt/v</td>
<td>Mean=2.6 SD=2.4</td>
<td></td>
</tr>
<tr>
<td>Mitton et al. (2000)</td>
<td>NSCP, Large D88</td>
<td>SCP</td>
<td>D88 (±0.2 mm)</td>
<td>D88</td>
<td>18 pt/v</td>
<td>Mean=2.6 SD=2.4</td>
</tr>
<tr>
<td>Mitton et al. (2001)</td>
<td>NSCP, Large D88</td>
<td>SCP</td>
<td>D88 (±0.2 mm)</td>
<td>D88</td>
<td>18 pt/v</td>
<td>Mean=2.6 SD=2.4</td>
</tr>
<tr>
<td>Mitulescu et al. (2001)</td>
<td>NSCP, Large D88</td>
<td>SCP</td>
<td>D88 (±0.2 mm)</td>
<td>D88</td>
<td>18 pt/v</td>
<td>Mean=2.6 SD=2.4</td>
</tr>
<tr>
<td>Mitulescu et al. (2002)</td>
<td>NSCP, Large D88</td>
<td>SCP</td>
<td>D88 (±0.2 mm)</td>
<td>D88</td>
<td>18 pt/v</td>
<td>Mean=2.6 SD=2.4</td>
</tr>
<tr>
<td>Bras et al. (2003)</td>
<td>NSCP, Large (cervical, thoracic, and lumbar vertebral)</td>
<td>SCP</td>
<td>D88 (±0.2 mm)</td>
<td>D88</td>
<td>18 pt/v</td>
<td>Mean=2.6 SD=2.4</td>
</tr>
<tr>
<td>Cheriet et al. (2007)</td>
<td>SCP, Small K07</td>
<td>SC9</td>
<td>D88 (±0.2 mm)</td>
<td>D88</td>
<td>18 pt/v</td>
<td>Mean=2.6 SD=2.4</td>
</tr>
</tbody>
</table>

*Paper not available. Data based on the abstract and in Novosad et al. (2004).

†Unknown/unreported large calibration device.

‡Geometrical model accuracy not evaluated, only clinical indexes. Some of these indexes are presented.

§Detected one incomplete vertebra in the validation scan. Real maximum error inside brackets.
Chapter 3

Thesis Statement

The main goal of this thesis is to develop an unsupervised method for constructing personalised geometrical models of scoliotic spines using two radiographs acquired in standard radiological environments. The key issues of this goal are:

- Unsupervised method: user intervention should be reduced to minimum and if any landmarks are required they should not depend on the number of vertebrae to reconstruct;

- Personalised model: the method should try to capture the features of the subject being analysed;

- Scoliotic spines: the method should be able to handle regular and scoliotic spines;

- Using radiographs: the method should use the standard examination that physicians prescribe to scoliotic patients - one PA (or AP) and one lateral radiographs;

- Standard radiological environment: the method is meant to work in ordinary radiological environments where no calibration object is present nor any kind of rotary platform, and therefore any adaptations to the environment or protocol should be minimum. This goes against the majority of the work in this area where a rigorous setup of the radiological environment is required. If we confirm that it is not possible to calibrate spine radiographs without calibration objects, we will try to minimise calibration equipment.

- Clinical application: the method should be accurate enough for clinical applications.
To our knowledge, currently, no method is able of tackling all these problems. In particular, we have not found any work that combines automatic model construction with standard radiological environments.
Chapter 4

Proposed Approach

In this chapter we will propose an approach to solve the previously discussed problems of spine radiography calibration, and construction of geometrical models of the spine using radiographs.

4.1 Calibration of spine radiographs

Radiography calibration is a requirement for accurate reconstructions of the spine. In the following sections we will propose methods for tackling the calibration problem where we will try to avoid calibration objects and minimise user intervention.

4.1.1 Avoiding calibration objects

Our first experiments will focus on trying to obtain a satisfactory calibration without using calibration objects that alter the content of radiographs. We believe that there is an alternative approach to using small calibration objects that passes by providing information that may be acquired during the examination. We will start by implementing a calibration method similar to Cheriet et al. (1999a); Cheriet and Meunier (1999); Kadoury et al. (2007a) and then we will try to improve it by using some parameters that may be acquired during the examination, such as, the horizontal distance from the x-ray source to the x-ray film plane and the height of the x-ray source. These parameters may be easily acquired by technicians when performing examinations using metric tapes or installing distance sensors in the x-ray machine. Distance sensors are possibly a better solution because of the lack of accuracy of metric tapes, even when compared with low cost sensors. However, the x-ray machine that will be used is equipped with a metric tape which enables to measure the distance to the x-ray
film without installing any other devices. We are also considering using other parameters such as the dimensions of radiographic films and direct measurements of the patient that may be easily acquired by the technicians. With the introduction of these parameters we intent to reduce the search space of solutions for the calibration parameters and consequently increase the probability of finding the global optimum solution. If we conclude that using this additional information is not enough to perform sufficiently accurate calibrations, we will try using radiopaque markers or small calibration objects that will appear in the radiographs. Such objects will have known dimensions and/or will be placed at known positions, which will enable to improve the calibration procedure. We will try to use standard and easily available equipment (e.g. calibration spheres from Xemarc - figure 4.1). In addition, we plan to experiment other optimising algorithms for least squares minimisation besides the Levenberg-Marquardt algorithm that was used in all the previous experiments.

Source: adapted from Xemarc (2007)

Figure 4.1: Radiopaque 25mm calibration balls from Xemarc.

Another adaptation that must be done is related to the position of the patient during the examinations. In most of the related work, the patient stands on a rotatory platform that enables to keep approximately the same distance between the spine and the x-ray source and between the spine and the film in both acquisitions (AP/PA and lateral). Additionally, it enabled to keep the same distance between the x-ray source and the film in both acquisitions, which facilitates solving the calibration equations. Moreover, the rotatory platform enabled to rotate the patient 90 degrees with high accuracy (±0.2 degrees) and without the patient having to move. Other approaches enable acquiring the two radiographs simultaneously by using a very specific x-ray machine. In our case, we will use the standard acquisition protocol for this kind of x-rays: first, the patient stands on the floor and places his body against the film support for the first acquisition, and then the patient rotates 90 degrees and places his shoulder against the film support for the second acquisition. This technique may originate complications for calibration proposes and may affect the accuracy of the geometric model construction because patients will have to move from one acquisition to the other and in the meanwhile change their posture. On the other hand, it will place patients closer to the film in both acquisitions, which will minimise the non-linear scaling effect of the diffuse x-ray source. To our knowledge, no work has been developed
that explores the standard patient positioning for this kind of radiographs.

4.1.2 Automatic calibration

For accurate calibrations a minimum number of landmarks is required, which in the current methods are manually introduced by a technician. In the work of Kadoury et al. (2007a) the authors estimated a minimum of 20 landmarks for best results. These landmarks are stereo-corresponding points (visible in both radiographs) and therefore are difficult to mark and subject to human errors. Additionally, these 20 landmarks are a subset that is chosen from the entire set of stereo-corresponding points that are used to construct the geometrical model of the spine. Therefore, in order to increase the quality of calibrations, more than 20 landmarks may be required for selecting the best ones and discarding the others. We propose using segmentation algorithms in order to automatically detect the most of the landmarks and consequently minimise user intervention. This procedure will try to automatically identify anatomical structures that are present in both radiographs, but it will probably need that technicians handmark some information first. We have already done some work in spine segmentation (Moura et al., 2006) where we implemented fully automatic algorithms for locating the spine and determining the box that encloses the vertebral body (figure 4.2). However we only analysed a small set of frontal radiographs and we did not use a priori information about the spine (and vertebrae) anatomy. The current state of the algorithm still lacks on robustness and is not prepared for severe scoliotic spines (for a complete description of the algorithm, please consult appendice A).

We plan improving segmentation by adding anthropometric information that we have been collecting about vertebrae shape and scoliosis effects on vertebrae shape, namely:

- Direct measurements of vertebrae (Bertrand et al., 2006; Tan et al., 2004, 2002; Berry et al., 1987; Panjabi et al., 1992, 1991);
- Vertebrae orientation along the spine (Gangnet et al., 2006);
- Statistical correlation studies between vertebrae measurements (Laporte et al., 2000; Lin, 2006);
- Quantitative analysis of vertebrae deformations on scoliotic spines (Parent et al., 2004b,a, 2002; Aubin et al., 1998).

We believe that with this information the algorithm will be able of making more informed decisions and consequently improve its robustness as well the segmentation quality. The output of the segmentation algorithm will be the box that encloses every vertebrae body. This box will provide information about
vertebrae position, dimensions, and orientation. We will use this information to determine stereo-corresponding points that will be used for calibrating the radiographs. Figure 4.3 illustrates the complete calibration process.

If required we may use Deformable Shape Models (DSM) for detecting specific vertebrae structures. This technique will be used for acquiring vertebrae shape and, therefore, it will be detailed in section 4.2.1, where we will introduce our approach for automatically constructing the geometrical model of the spine.

4.1.3 Validation

For validating the proposed calibration method we will use a set of radio-opaque objects (e.g. cubes, pyramids, cylinders) with known dimensions. These objects will be placed at a given position with a given orientation. Two radiographs will then take place: the first with the objects in the initial position and the second after rotating them 90 degrees (simulating the effect of patients rotation). After this, a set of stereo-corresponding points (SCPs) of the objects will be marked on both radiographs. We will then run the proposed calibration algorithm for determining the calibration parameters. Using the SCPs and the calibration parameters, the objects dimensions will be calculated and compared with their real dimensions. This test will be repeated several times with different parameters (e.g. x-ray source distance to the film and height from the floor, objects positions
4.1. CALIBRATION OF SPINE RADIOGRAPHS

Validation of the automatic calibration process will be performed on radiographs where a spine and radiopaque objects of known dimensions are both present. This will enable to compute calibration parameters using information about the spine and then, using these parameters, the dimensions of the object of known dimensions may be calculated. Calibration errors will be estimated comparing the objects’ calculated dimensions with their real dimensions. Furthermore, a manual calibration may be performed (using handmarked stereocorrespondent points on vertebrae) for comparing results with the automatic method and determine which is the most accurate. The main difficulty of this validation procedure is introducing objects of known dimensions on in vivo examinations because these objects must be subject to the same geometrical transformations that the patient is when rotates from one radiography to the other. This obliges keeping these objects together with the patient, having special carefull to not jeopardise the radiographs content. An alternative would be using dried vertebrae instead of real patients but the testing conditions would be ideal for the segmentation algorithm and, therefore, one should always perform an in vivo validation. Finally, we plan to estimate the minimum information (e.g. landmarks) that a user should manually introduce to achieve best results.
4.2 Computerised modeling of the spine

After calibrating radiographs, the next step is to capture the spine geometry. In the following sections, we propose methods for performing this task automatically and procedures for validating these methods.

4.2.1 Unsupervised construction of geometrical models

We believe that for constructing geometrical models of the spine, the content of radiographs should be fully explored. In other words, the reconstruction should try to use most of the information available in radiographs about vertebrae shape (e.g. contours or silhouettes) instead of being limited to a set of landmarks, like it happens in most of the related work. Additionally, it is our goal developing an unsupervised method, meaning that user intervention should be minimum or nonexistent. For coping with these issues, we are considering utilising Deformable Shape Models (DSM). DSMs usually use generic models (atlas) that are deformed and transformed in order to fit to the image content. These deformations are guided by an optimisation procedure that starts from an initial solution, which in our case will be a generic model of a vertebra. Thus, DSM will be applied in a vertebra by vertebra basis. For trying to give a good initial estimation of vertebrae shape, position and orientation, we will use the segmentation results from the automatic calibration, in particular:

- The location of the vertebra in both images;
- The vertebra level (e.g. L3 or T6), which will enable to select an atlas for that specific vertebra level;
- Sagital and coronal rotation of the vertebra;
- An estimation of the axial rotation using the position of vertebrae pedicles (if available);

We believe that with these hints the DSM will have a higher probability of finding a good solution. Let us remember that not having a good initial solution was one of the flaws of Benamour et al. (2003). In figure 4.4 we illustrate the complete process of geometrical modeling of the spine.

For constructing generic models (atlas) three options are available: (i) using a generic spine from one specimen (one vertebra per level), (ii) using several vertebrae per level from different specimen to build a mean model, and (iii) adding to the mean model standard deviations. The first option is more limited but also the easier to execute because it only requires the construction of one model per vertebra level. The other two are able to capture a more representative model of every vertebra. In particular, the third option enables capturing statistical information about deformations, which would be very useful if we had access to both
4.2. COMPUTERISED MODELING OF THE SPINE

Figure 4.4: Automatic vertebrae shape modeling.

healthy and scoliotic vertebrae. This last option was used by Benamour et al. 
(2001, 2003, 2005) and, in our option, is the one with higher potential for capturing vertebrae shape. However, it needs a representative set of vertebrae models that may be difficult to acquire in the timespan of this thesis.

Generic models will be constructed from CT scans, which are able of capturing vertebrae geometry with high detail. We already started experimenting techniques for retrieving the geometrical model of each vertebra from a CT scan. Probably, the method for constructing the model for one vertebra will have the following steps:

1. Applying a segmentation and a thresholding algorithm (already available in CT software) for removing the structures from the CT that are not bone tissue;
2. Extracting the CT slices for the given vertebra;
3. Removing the bone tissue from adjacent vertebrae and ribs;
4. Applying a marching cubes algorithm (already available in CT software) for constructing the geometrical model of the vertebrae;
5. Marking the landmarks that will be used by the DSM.

Currently this process is very manual. However, for constructing several models per vertebra level we may need to automate this technique. The last step is
critical because the landmarks must be the same for all vertebrae from the same level, in other words, they must be placed in the same anatomic regions respecting a predefined order for enabling matching landmarks between models. Moreover, the number of landmarks must be sufficient to produce morpho-realistic models.

In conclusion, we will use DSM for extracting vertebrae shape and we will try to use a mean model with information about probable deformations. For giving a good initial solution to the DSM we will use the segmentation results from the automic calibration procedure. A literature review on DSM will be required for selecting the best approach.

4.2.2 Validation

Both qualitative and quantitative validations will take place. For qualitative validation we propose superimposing reconstructed models with radiographs and with CT reconstructions. During CT scans the spine is in a different posture than in radiographs and, therefore, only the vertebrae shape will be validated, vertebra by vertebra. On the other hand, on radiographs the complete reconstructed spine will be superimposed. This qualitative techniques will enable physicians to visually evaluate reconstructions. For quantitative validation we will measure the difference between ground-truth models and the reconstructed models in two ways: point-to-surface distance (distance between every point of the reconstructed model and the nearest polygon of the ground-truth model) and volumetric comparison (volume percentage in common between the reconstructed model and the ground-truth model). Currently, we are considering the following options:

- Using radiographs and ground-truth models from other studies: contacts are being made for testing our algorithms in the same dataset that other studies use, which will enable a direct comparison with the other methods.

- Obtaining radiographs and CT scans for the same patients: this would enable comparing reconstructions from our algorithm against reconstructions from CT. However, it is not usual for a patient to do both examinations, which will difficult collecting a significant number of examinations. Using cadavers it is also difficult because of logistic and administrative reasons.

- Simulating radiographs from CT scans: it is possible to construct simulated radiographs from CT scans, which would solve the previous issue. However, the method that will be developed will have as input real radiographs, and therefore, it should be validated with them. Nevertheless, if none of the previous options will be possible, we will use simulated radiographs.

The validation procedure is highly dependent of the examinations that we will get access to. At the moment, the only validations we are able to perform are
(i) qualitative evaluation by superimposing reconstructions over radiographs, and
(ii) quantitative evaluation using TC reconstructions as ground-truth model and
using simulated radiographs (from the same TC scans) as input for our algorithm.

4.3 Summary

In this chapter we presented the approaches that will be used for spine radiography calibration and construction of personalised models of the spine using radiographs. For the calibration problem we propose avoiding calibration objects and using extra information that may be acquired during examinations, such as the x-ray source position. We also proposed automating the calibration process in order to reduce user intervention. For this task we will improve segmentation algorithms that we already started developing. Validation of calibration will be made using radiopaque objects of known dimensions.

Construction of geometrical models will use Deformable Shape Models (DSM), which try to deform generic objects for best fitting the image content. We will use the segmentation results from the automatic calibration for positioning generic objects (vertebrae) and thus constructing a good initial solution of the spine model. Generic objects will be constructed from CT scans and we will try to collect several vertebrae per level in order to have mean vertebrae models and typical deformations. The validation procedure highly depends of the examinations that we will get access to. Basically, both qualitative and quantitative validations will be done, the first by superimposing reconstructed models on examinations for visual evaluation, and the second by measuring differences between reconstructed models and ground-truth models (typically constructed from CT) of the same specimen.
Chapter 5

Work Plan

In this chapter we start with a brief description of the previous work. After that, we present the work plan for the next three years of this thesis.

5.1 Previous work

It is important to mention that during this last year some work was already developed. During 6 months a very simple tool was built for reconstructing the spine where we had the opportunity to get in touch with the problem. In this context, we implemented some segmentation algorithms that may be used for this thesis. It is important to mention that this tool is very basic because it only tries to capture the spine curvature without trying to mimic vertebrae shape and axial rotation. Additionally, no calibration method was used, which obviously compromises results.

We also reviewed and collected several literature concerning radiological images calibration (presented in section 2.1), geometric models construction from radiographs (presented in section 2.2), construction of physical models, medical image segmentation, spine anatomy, anthropometry of the spine, and scoliosis and its morphological alterations on vertebrae. This process also took about 6 months, but we are constantly updating our references database. Moreover, in parallel with the literature review, we have been experimenting modelling tools, which will be used later for constructing geometrical models of generic vertebrae.

5.2 Work plan for the following years

Here we present the work plan for the following years: first at a broader view in table 5.1 and then with a more detailed explanation task by task.
Table 5.1: Work plan for the next three years.

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<tr>
<th>#</th>
<th>Task</th>
<th>Year</th>
<th>Trimester</th>
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<td>1</td>
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<td>3</td>
<td>Automatic Calibration</td>
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<td>4</td>
<td>Atlas Construction</td>
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<td>5</td>
<td>Automatic Atlas Deformation</td>
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<td>6</td>
<td>Dissertation Writing</td>
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1. Manual Calibration: Implementation of algorithms for calibrating radiographs without calibration objects, using landmarks handmarked by a user. Optimisation algorithms will be implemented for finding the optimum calibration parameters. Additionally, the implemented algorithms will be validated and compared with existent work. This task may have some time gaps especially due to eventual equipment acquisitions/manufacturing and because of experiments that must be performed in external radiologic laboratories. The time spent with this task will vary with the success of overcoming the challenges of minimising both calibration equipment and changes in the radiography acquisition protocols.

2. Spine Segmentation: This task is independent of the manual calibration, during which some time will be available for initiating experiments with radiography images segmentation. This task will start with a more exhaustive review on radiography segmentation and then will continue to the following subtasks: spine midline detection, vertebral bodies segmentation, and automatic identification of other vertebral structures. This information will be used in the next task.

3. Automatic Calibration: After manual calibration is accomplished, automatic calibration will be experimented using the calibration algorithms already developed in the first task, and using the output of the segmentation algorithms. There will be a strong interaction with the segmentation task because automatic calibration depends on the segmentation results. Validation of calibration results and comparison with other works is also included here. Once again, the success of the attempts on using automatically detected data for calibrating radiographs will greatly influence the duration of this task.

4. Atlas Construction: This task is concerned with the construction of a generic model of the spine, which includes models of vertebrae. Such model will be constructed using CT scans, and will be used in the next phase. The timespan of this task will depend on the type of atlas used, as discussed in section 4.2.1.
5. Automatic Atlas Deformation: While executing the previous task, a broader review of Deformable Shape Models will be performed. After completing the previous task, a generic model of the spine will be available for experiment some of the reviewed Deformable Shape Models with eventual variations. Such algorithms will be used for fitting the generic model of the spine in the radiographs of individuals, which will produce personalised geometric models. This task also includes validating the experimented approaches and comparing results with other works.

6. Dissertation Writting: We reserve about six months for writting and reviewing this thesis dissertation.
Chapter 6

Implications of Research

There are several approaches in the literature for constructing geometrical models of the spine using two orthogonal radiographs as input. However, most of the methods need very special equipment for calibrating the radiological environment. Additionally, spine reconstruction from radiographs is still very manual, which makes of it a time-consuming and resource-consuming task. We intend to create a method that may be easily adapted to any radiological environment and that will require minimum user intervention. Such method, if sufficiently accurate, will enable to perform a better diagnosis of scoliosis, which is a 3D deformation of the spine and, therefore, difficult to evaluate using plain radiographs. Additionally, by offering a better understanding of the patients’ spine deformation, physicians will be able to perform more informed decisions without having to expose the patient to extra examinations and extra radiation. Nowadays, there are few hospitals that are well equipped and already benefit from such advantages, but by automating spine reconstruction and by trying to minimise changes in standard radiological environments and protocols, these advantages may be available to many more health institutions.

Besides these implications, during this doctoral program we expect to contribute to the areas of medical image analysis by providing algorithms for locating vertebrae position, assessing vertebrae rotation, and for vertebrae segmentation and registration. Such methods may be adapted later to address other problems besides spine model construction, such as, detection of vertebrae pathologies.

Finally, we hope that at the end of this thesis we have opened the doors for start exploring personalised biomechanical simulations in our laboratories. This kind of simulations would be of great assistance for planning and prescribing therapy for scoliotic patients. Furthermore, they would help designing new corrective techniques.


Appendix A

Paper on vertebrae segmentation

Paper presented on CompIMAGE06 (Moura et al., 2006) describing preliminary work related to this thesis.
Automatic Vertebra Detection in X-Ray Images

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ABSTRACT: In this paper we will describe our experiments with x-ray image analysis for vertebra detection in juvenile/adolescent patients with idiopathic scoliotic spines. We will focus on detecting vertebrae location in a fully automatic way. For accomplishing this, we propose a set of techniques for (i) isolating the spine by removing other bone structures (e.g. ribs), (ii) detecting vertebrae location along the spine using an hierarchical and progressive threshold analysis, and (iii) detecting vertebrae lateral boundaries.

1 INTRODUCTION
In this paper we will describe our experiments with x-ray image analysis for vertebra detection. The input of the image analysis process is a pair of high-resolution grayscale images obtained by x-ray examinations. Both images are obtained at the same time and capture the spine of a given person, although from different perspectives: anterior-posterior (Fig. 1) and lateral (Fig. 2).

Our goal is to detect the 3D geometric location of each vertebra for constructing a 3D model of the spine. We intend to accomplish this by using a common and affordable diagnosis examination such as x-ray images. The main objective is to enable physicians to visualise in a 3D perspective a model that approximates the spine of their patients, obtained after processing of standard diagnosis examinations already available. Additionally, the vertebrae detection process should be able to deal with examinations of juvenile/adolescent patients with idiopathic scoliosis (condition that involves an abnormal side-to-side curvature of the spine). However, the methods presented here are not yet prepared for handling severe conditions of scoliosis.

Some work has been developed in the area of vertebra detection. A considerable part of the methods developed for vertebra segmentation in x-ray images need a set of images manually labeled by experts. These images are used as training sets for the program to build a model of the vertebrae. This model is then used to determine vertebrae location and form in other images (de Bruijne and Nielsen 2004b; de Bruijne and Nielsen 2004a; Zamora, Sari-Sarrafa, and Long 2003; Smyth, Taylor, and Adams 1999; Scott, Cootes, and Taylor 2003). Additionally, research is being developed in different types of examinations, such as, DXA scans (Smyth, Taylor, and Adams 1999; Scott, Cootes, and Taylor 2003) or CT scans (Ghebreab and Smeulders 2004). Apart from that, the authors usually choose to segment a small set of vertebrae, like the lumbar or the cervical. Our work differs from the previous, since we use x-ray images only, we intended to detect the location of the maximum number of vertebrae possible, and we do not need to have access to images labelled by experts. Using these same principles, Benameur and Pomero already were able to achieve interesting results. Benameur was able to reconstruct the lumbar and part of the thoracic spine using frontal and lateral x-rays with little user intervention (Benameur, Mignotte, Labelle, and Guise 2005). The user has to mark two landmarks in one of the vertebrae in both x-rays and the rest of the process is automatic. Pomero accomplished to reconstruct the vertebral body from T1 to L5 with a semi-automated
method (Pomero, Mitton, Laporte, de Guise b, and Skalli 2004). The user has to identify the corners of every vertebra in both lateral and front projections. However, this process may be speeded-up using an algorithm that automatically calculates part of the landmarks based on the landmarks already marked by the user.

The method we propose here, tries to minimise user interaction. We are aiming for detecting the most of the vertebral body without the user having to insert any landmark in order to eliminate possible user related errors. For accomplishing this, we will start by analysing the anterior-posterior (AP) image. As we can see in Figures 1 and 2, the AP projection is much richer in information than the lateral projection, in it one is able to see and identify almost every vertebra. On the other hand, in the lateral projection the rib cage and the arms make it difficult to identify vertebrae, even for a human expert. Therefore, our strategy is to start by analysing the AP image to obtain the X and Y coordinates. We will then try to obtain the vertebrae depth (Z coordinate) using information about the curvature of the exterior boundary in the lateral projection.

In the next sections we will focus on the analysis of the front perspective image and we will demonstrate how we were able to determine vertebrae positions in the XY plane.

2 DETECTING VERTEBRA POSITIONS IN THE FRONT PERSPECTIVE

The x-ray images that feed the image analysis process have much more information besides the spine. There are a lot of bones structures present in the image that we do not need and that may difficult detecting vertebrae. Therefore, the first step consists in isolating the spine in order to remove undesired information, such as, the ribs, the head and legs. After isolating the spine we determine where each vertebra begins and ends along the spine (along the Y axis). For doing this, we "walk" through the spine searching for discontinuities that may indicate vertebrae limits. Finally, having identified the Y limits of every vertebra, we are able to determine their lateral limits using local information. In the next subsections, we will describe the process of vertebra detection in detail by addressing the issues of spine isolation and limits detection separately.

2.1 Isolating the spine

Isolating the spine is a crucial step in the process of vertebra detection. The main goal is to remove the major bone structures that are not vertebrae and whose presence may cause errors in the following steps.

Currently, we are working with high resolution images (2448x3264) with very high detail but considerable noise. For detecting a big object, such as the spine, we built a multiresolution Gaussian pyramid. This pyramid allow us to analyse the image at lower
resolutions and therefore with less detail. Small objects disappear and the image is more regular, which facilitates detecting the spine boundaries. In Figure 3 we present the result of downsampling the original image 5 times.

In order to isolate the spine, we first have to know its location. As we can see in the previous figures, the image columns with more bright pixels are the columns where the spine is located. This happens because bright and large bone structures like the spine, head and hips are vertically aligned. Therefore, for determining the spine location in the X axis, we start by counting the number of bright pixels in every column and then we select the column with the highest counting. In spite of the simplicity of this technique, it turned out to be very robust in the available sample of images. Figure 3 shows a graphic of the column count (at the bottom) and a vertical line representing the selected centre in the X axis.

Next, we try to isolate the head, spine and legs, removing all other bone structures (e.g. ribs, clavicles, arms). For achieving this, we start by thresholding the image to remove the objects with low intensity. Usually, vertebrae have the pixels with highest intensity in their neighbourhood, although this brightness varies a lot among them. Therefore, the threshold is done locally, line by line. This operation is able to isolate part of the spine, but fails in areas where other bone structures have more intensity than the vertebrae. To solve this problem we analyse the image from the bottom to the top (legs to head) and we monitor the width of the body. From the moment that the width starts to decrease (more or less at the hips) we constrain its increase. This will allow to ignore ribs and other structures that are considerably wider. We also control the centre evolution to prevent deviations. Of course, with this constrain the head "becomes" very narrow, which difficults differentiating it from the spine. To avoid this, we repeat the same process, but now from the top to the bottom (head to legs) and we merge the two results. The result is presented in Figure 4a (after using morphing operations for closing some holes). Finally, we obtain the image illustrated in Figure 4b by upsampling the mask and performing a bitwise AND operation with the original image.

Removing the head, hip and legs is then accomplished by detecting large concentration of bright pixels in the top and bottom of the image respectively.

### 2.2 Detecting vertebrae limits in the Y axis

Having isolated the spine, the next task is to divide it in vertebrae. If we look closely, we can see that vertebrae are usually bright and the disks that separate them have lower intensity. Based on this observation, we built an algorithm that detects discontinuities along the spine and tries to figure out if they may indicate the presence of a disk separating vertebrae. How-

![Figure 3: Body center detection](image)

![Figure 4: Spine isolation. Isolation mask (a) and the result of applying the mask to the original image (b).](image)
ever, vertebrae intensity vary a lot: cervical vertebrae usually have low intensity and lumbar vertebrae usually present very high intensity. This makes it difficult to classify regions as vertebrae because we cannot define pattern levels of intensity. The intensity of a vertebra depends on its position and of the image acquisition equipment. For tackling this problem our algorithm uses a progressive thresholding approach. The algorithm starts by counting the number of pixels per row at a very low threshold. Then, the threshold value is incremented at a slow rate and the counting process is repeated. Figure 5 illustrates in the right side the result of applying this technique, although we only included some threshold values for demonstration proposes. As we may observe, with low threshold levels we are able to isolate vertebrae with low intensity (typically at the cervical) and with higher threshold levels we accomplish to detect vertebrae with higher intensity.

Figure 5: Detecting nodes along the spine (thresholds from left to right: 32, 48, 176, and 192)

Nevertheless, this algorithm has two issues that must be solved in order to correctly divide the spine into vertebrae: (i) vertebrae may become shorter and shorter while incrementing the threshold, and (ii) vertebrae may be divided in several smaller regions due to intensity variations along vertebrae. For handling these problems we decided to use a tree data structure to store the regions that the algorithm detects. Every time a new region is found, it is added to the tree as a child of the smallest region that entirely encloses the new region. In order to control the tree size and to overcome the problem (i), before increasing the threshold to search for new regions, we prune the tree by removing the leaves which have no siblings. Leaves with no siblings are not interesting because they do not divide the parent region. At most, they reduce the size of the parent region which is not interesting because vertebrae should have maximum height in order to stay close to each other. By the end of the algorithm, the tree is fully constructed and its leaves should represent vertebrae, unless some vertebrae were over-divided. For detecting over-divided vertebrae we do two tests: (i) we check if the gaps between vertebrae are not too large, and (ii) we determine if the vertebra size is consistent with its adjacent vertebrae (e.g. if it is not too small compared to its largest adjacent vertebra). Whenever one of the previous situations is detected, we test if the leaf’s parent is a better candidate for that vertebra. If that is the case, we remove all the leaf’s parent childles, transforming the parent into a leaf and therefore in a vertebra.

2.3 Detecting vertebrae limits in the X axis

After detecting where the vertebrae are located along the spine, we must detect where they start and end along the X axis. This operation may be more difficult than what it seems because part of the ribs may still be attached to vertebrae in the processed image. This happens when ribs also show high intensity levels and the spine isolation method is not precise enough to get rid of them.

For detecting vertebrae X limits we divide them in several clusters along its width (currently we divide it in 15 clusters). Then, we rank the clusters according to their intensity levels. Intensity levels are calculated using an exponential scale to give more preponderance to very high levels. This allow us to distinguish between clusters with high intensity structures surrounded by low intensity pixels, and more homogeneous clusters with average intensity levels. We then select the first three clusters with more intensity and we elect them as candidates for being the X limits. One of these candidates will represent the start of the vertebra and the other will represent the end. Initially, the two more intense clusters are selected. Then, for each vertebra, we compare its width and X centre with its nearest 4 vertebrae. If we detect a considerable deviation of the vertebra centre or an unexpected change in width, we try different combinations of the three candidate clusters and we select the ones that best fit the conditions.

Finally, we optimise the results by finding inside the elected clusters the largest concentration of bright pixels. Only then the process of detecting the X limits is completed. In Figure 6 we may see the nodes
fully identified with the centre marked with red small squares.

The next step would be detecting the Z coordinate of each vertebra. For accomplishing this, we intend to use the body curvature that is observable in the lateral perspective (Fig. 2) and the already calculated Y coordinate, which tell us where to find each vertebra along the spine.

3 CONCLUSIONS

In this paper, we have proposed a set of techniques for detecting the vertebrae location in X-ray images in a fully automatic way. We started by isolating the spine for removing other bones structures. We then used a progressive thresholding algorithm for detecting vertebrae along the spine, which uses a tree data structure to store regions that may correspond to vertebrae. After pruning the tree, its leafs have the vertebrae location in the Y axis. Finally, the X boundaries of each vertebra is determined by performing an intensity analysis along the vertebra width.

So far, we have obtained promising results for detecting vertebrae in the anterior-posterior projection.

Our next step is to improve the present process using domain specific information, such as, a spine model. We will then try to detect vertebrae location in the lateral projection, and use all captured features to produce a 3D model of the spine.

References


